

ABSTRACT

Title of dissertation: THE DYNAMICS OF MULTI-SENSORY RE-WEIGHTING
IN HEALTHY AND FALL-PRONE OLDER ADULTS

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Multi-sensory re-weighting (MSR) is an adaptive process that prioritizes the visual, vestibular and somatosensory inputs that provide the most reliable information when environmental conditions change. This process is thought to degrade with increasing age, and to be particularly deficient in fall-prone versus healthy older adults. This dissertation presents three studies designed to investigate age- and fall-related changes in MSR.

The first study examined the assumption of impaired MSR in healthy and fall-prone older adults using a two-frequency touch/vision experimental design with stimuli at varying amplitudes. Both healthy and fall-prone older adults demonstrated the same pattern of adaptive gain changes as healthy young adults. No group differences in the overall levels of vision and touch gain were found. These results suggest that, for small amplitude vision and touch stimuli, the central sensory re-weighting adaptation process remains intact in healthy and fall-prone older adults.

In the second study the effects of a sensory-challenge balance exercise program on laboratory measures of MSR and clinical measures of balance were investigated. Following the intervention the normal adaptive pattern of gain change was unaltered, while declines in overall vision and touch gains that reflect down-weighting of the sensory stimuli were seen. Improvements in four clinical balance measures were observed. These findings indicate that MSR processes in fall-prone older adults are modifiable, that sensory challenge balance exercises may facilitate the ability to down-weight unstable sensory inputs, and that these effects may generalize to other components of balance.

A third study explored the dynamics of sensory re-weighting in healthy and fall-prone older adults. Absolute levels of gain, and the rate of adaptive gain change, were examined before and after large changes in visual motion stimulus amplitude. Compared to young adults, gains in both older adult groups were higher when the stimulus amplitude was high, and gains in the fall-prone elderly were higher than both other groups when the stimulus amplitude was low. Both older groups demonstrated slowed sensory re-weighting over prolonged time periods when the stimulus amplitude was high. These results reflect age- and fall-related changes in the extent and rate of down-weighting unstable visual inputs.

**THE DYNAMICS OF MULTI-SENSORY RE-WEIGHTING
IN HEALTHY AND FALL-PRONE OLDER ADULTS**

By

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Dissertation submitted to the Faculty of the Graduate School of the
University of Maryland, College Park, in partial fulfillment
of the requirements for the degree of
Doctor of Philosophy
2006

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Dedication

To Joshua,
who lives in my heart

To Howard and Shirley,
with gratitude and my commitment to the enhancement of successful aging
for all older adults

And to David,
who knows swimology
and kept me afloat

Acknowledgements

I offer my warmest thanks to my friends and colleagues in physical therapy for their support and encouragement, and to my fellow graduate students in the Cognitive Neuroscience Motor Behavior Lab for their humor, tolerance and valuable assistance.

I extend my sincere appreciation to the members of my committee for their guidance and studied advice. I am especially indebted to Tim Kiemel for his patience and precision, and to Fay Horak for two decades of mentorship and inspiration.

Lastly, I am deeply grateful to John Jeka for teaching me the science of science and the art of thinking, and persevering with me to achieve this lifelong goal.

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Chapter 1: A Review of the Literature

Introduction

The problem of falls in the elderly has emerged as a serious health threat to older adults and a significant drain on the limited pool of health care resources. Increased attention to this problem has resulted in a growing body of evidence on the causes and consequences of falls, and improved clinical methods of evaluation and treatment for older adults who fall. Imbalance is recognized and accepted as a major cause of falls, but what are the causes of imbalance in fall-prone older adults? To answer that question, an understanding of the components of the postural control system and age-related changes within the system is necessary.

Balance is a multidimensional process that relies upon peripheral reception of information from the environment, central perception of body and spatial orientation, central response selection and planning, and peripheral execution of automatic and volitional motor responses (Horak, Shupert et al., 1989). Perception and action are inseparable: the generation of balance responses that are appropriate to environmental conditions depends upon an accurate sense of orientation within that environment.

Separate somatosensory, visual and vestibular inputs comprise the primary sources of information that contribute to postural orientation. The process of melding these three distinct senses into a single unified perception of body-in-space is “sensory integration”. The mechanisms by which the central nervous system accomplishes this process are not well understood (Anderson RA, Snyder LH et al., 1997).

For non-impaired persons under usual environmental conditions, information from the three senses is partially redundant and the determination of body position in space is automatic and routine. In certain environments, however, information from one or more sensory channels may be absent, or altered such that conflict between the senses arises. When this occurs, central mechanisms must adjust to a reduction in sensory information or resolve a sensory conflict by re-weighting incoming sensory information to rely more heavily on the available and accurate sensory sources (Horak, Shupert, and Mirka, 1989; McCollum, Shupert et al., 1996). For example, if the surface conditions are firm and stable, proprioceptive inputs from the feet and ankles are more relevant than when standing on a soft and compliant surface; in the latter case visual and vestibular inputs become more important than usual because the amount of accurate, available information from the proprioceptive system is reduced. Central sensory selection and re-weighting processes appear to degrade with age, and may be particularly deficient in the fall-prone elderly (Alexander, 1994; Horak, Shupert, and Mirka, 1989; Woollacott, 2000).

This chapter provides a broad overview of the literature relevant to postural control in healthy and fall prone older adults. In a recent study of multi-sensory re-weighting in healthy and fall-prone older adults, using improved experimental design and methods, we have found evidence contrary to the established notion of deficient multi-sensory re-weighting in the elderly; this study is discussed in Chapter Two (Allison et al, in press). Subsequent to that investigation we undertook an intervention study to explore the effects of a sensory-challenge balance exercise program on multi-sensory re-weighting and clinical measures of balance in a group of fall-prone older adults (Allison

et al, submitted). This study, which offered additional evidence for intact and modifiable re-weighting in fall-prone older adults, is presented in Chapter Three.

How may we explain the discrepancies between our results, indicating relatively preserved re-weighting in fall prone elders, and the results of several prior studies that found deficient re-weighting in this same population? Differences in our experimental design, screening process, and measurement methodology may have led to our contrary results. In particular, most prior studies used short measurement trials (typically 10 to 30 seconds), while our trials were two minutes in duration. It may be that, *given enough time*, fall-prone older adults can successfully re-weight sensory information, but are unable to do so as quickly as young adults. Evidence from a very few studies of the dynamics of sensory re-weighting in young adults indicate that the time scale for the completion of adaptation may be on the order of five to eight seconds (Dijkstra, Schoner et al., 1994a; Peterka and Loughlin, 2004) and in some young subjects as much as 18 seconds. If, hypothetically, fall-prone older adults require twice as much time to fully re-weight, then in previous studies with short trials they may have been unable to demonstrate their actual re-weighting capabilities. A third study to compare the time-scale of re-weighting in healthy young, healthy older and fall-prone older adults is described in Chapter Four. There we demonstrate that multi-sensory re-weighting is not absent, but rather slowed, in fall-prone older adults.

Aging and falls

Dramatic improvements in life expectancy within the past century have led to an enormous increase in the proportion of the United States population that is 65 years of age or older (Rowe JW and Kahn RL, 1998). Today, more than 13% percent of U.S. residents are over 65, up from only 4% at the turn of the last century. In absolute numbers, that equates to over 33 million 'seniors'. The fastest-growing population group is the 'old-old', persons age 75 years and older. The number of older adults expected to live to age 100 or more is expected to rise nearly 10 fold in the next 50 years, from about 65,000 today to more than 600,000 in 2050. Women continue to outlive men; 75% of centenarians are women.

Falls are a serious health problem for older adults, with significant personal and societal ramifications. Naturally, as the aged population increases, so does the number of falls and fall-related injuries, disability and death. Reducing the incidence of falls in the elderly is critical to improving the lives of older Americans and to controlling their health care costs. Substantial efforts are being made by the scientific and health care communities to address this issue.

Falls are prevalent, with one-third of persons age 65 and older experiencing at least one fall annually (Tinetti, Speechley et al., 1988). In persons over 80 years of age, the incidence of falls is estimated at 40% to 50%. Half of elderly persons who fall each year do so repeatedly (Tinetti, 1989). Given the population statistics above, we can estimate that there are currently in excess of 5.5 million repeat fallers, the group most prone to injury and future falls. There are more falls among women than men, and more

falls among institutionalized elders versus those who live in the community (Sattin, 1992).

Falls have serious negative health sequelae for older adults, as well as sizeable economic consequences for families and society. Falls are the leading cause of injury and hospitalization due to injury in those over age 65; they are the most common cause of accidental death in persons over age 80 (Sattin, 1992). From 10% to 15% of falls result in serious injury requiring medical attention, with an additional 30% to 50% resulting in minor injury (Tinetti, 1989). Fractures occur in 5% to 10% of falls, and hip fractures in 1% to 2% of falls. There are now in excess of 300,000 hip fractures annually in the U.S. Of those who sustain a hip fracture, one in ten will die from complications, and 20% to 30% will remain institutionalized one year later (Nevitt MC, 1997). One in four hip fracture survivors will never regain their previous level of mobility and function (Tinetti, 1989). Injuries are more common among women than among men at all ages. Falls are also associated with loss of function and disability in the elderly, and often lead to a downward spiral of reduced activity, decreased mobility, isolation and loss of independence (Tinetti, Speechley, and Ginter, 1988). Among older adults who fall, irrespective of injury, 25% reduce their activity level immediately, and one year later 16% are still less active than before the fall. Falls lead to financial burdens for families and society. The direct costs for treatment of hip fractures alone now exceeds 10 billion dollars per year (Rizzo et al, 1998).

The high incidence of falls in the elderly combined with an earlier view of aging as a process of inevitable, irreversible and linear or curvilinear decline (Woollacott and Shumway-Cook, 1990;Woollacott, 1993), led to the assumption that poor balance and

falls were an inescapable fact of life for the elderly. This assumption appeared to be supported by several early studies of balance in older adults that in hindsight failed to adequately screen aged participants for pathology or physical impairments. The importance of such screening was shown by Gabell & Nayak (1984) , who carefully screened a large number of older adults and then tested a small subset of participants without detectable pathology or impairment. They found that compared to young subjects, postural control in these disease- and impairment-free older adults was only slightly diminished. Subsequent studies that employed more careful screening methods have confirmed that healthy older adults have mild declines in balance but do not have functional limitations or falls (Whipple, Wolfson et al., 1987). It is now well accepted that falls are not a normal part of aging. After all, two of three older adults between the ages of 65 and 80 years do not fall each year, nor do 50% to 60% of those over age 80. Falls affect only a portion of the elderly and are almost always associated with the presence of pathology or impairments (Studenski, Chandler et al., 1991;Horak, Shupert, and Mirka, 1989).

These findings are consistent with a more recent, competing model of the aging process that posits relatively stable performance over the course of the lifespan, with periodic performance declines caused by pathology or injury. In this model, frailty, loss of function and falls only occur after an accumulation of pathologies/injuries sufficient to reduce performance below a minimum functional threshold. Interventions that prevent or remediate pathology and injury, including lifestyle changes and exercise, are thus able to minimize frailty, loss of function and falls. Consistent with this ‘modifiable aging’ model are studies that show the positive effects of activity and exercise for the improvement of

balance and reduction of falls in older adults (Overstall, 2004). We will discuss several of these studies later.

Before we may confidently state that imbalance and falls are not due to age alone, but rather to pathology/injury, we must confront the statistic that just over 50% of all ‘fallers’ have no known medical diagnosis that would explain their falls, e.g., stroke, Parkinson’s Disease, peripheral neuropathy, etc. It is thought that many of these fall-prone older adults have subtle pathologies in several of the components of the postural control system, none of which are severe enough within any one component to earn the individual a medical diagnosis, but which in combination negatively alter postural control (Horak, Shupert, and Mirka, 1989). Within the fields of aging and gerontology, the concepts of ‘successful aging’ versus ‘usual aging’ have become prominent (Rowe JW and Kahn RL, 1998). The small group of ‘super senior’ subjects tested by Gabell and Nayak (1984) were likely ‘successful’ agers, that is, healthy, fit and active without discernable pathology or impairment. Older adults in the ‘usual aging’ category are likely to have some pathologies or impairments that may put them at risk for disease or disability, but not so severe or numerous as to prevent successful adaptation to preserve functional independence. Older adults with significant imbalance who are experiencing falls and functional decline could be termed ‘unsuccessful agers’. Some of these older adults will have known pathology and hence one or more medical diagnosis, but others without such a diagnosis will likely still have detectable impairments affecting multiple components of the postural control system.

In our studies we have recruited both older adults who fall (‘fall-prone’) and older adults who are healthy, fit and active with no history of falls (‘healthy’). All subjects in

both groups are carefully screened and free of known diagnosis that might impact balance, e.g., stroke, diabetes, etc., as well as free of any major impairments in the specific components of the postural control system that would affect their ability to perform our experiments, e.g., macular degeneration, etc. All subjects in both older groups are clinically screened for peripheral sensory loss and are excluded if such loss is present. In this regard, our fall-prone subjects may be atypical from their ‘unsuccessful aging’ peers.

The relationship between balance and falls

With the growing acceptance of the idea that falls were not explained by age alone, Yale geriatrician Mary Tinetti pronounced that falling was “a clinical entity in its’ own right” (Tinetti, 1989; Tinetti and Speechley, 1989) and therefore deserving of focused scientific attention and medical intervention. Increased research into the causes of falls has produced a consensus that falls in the elderly are the result of multiple risk factors. Risk factors can be categorized as intrinsic (associated with the aging individual, e.g., weakness, sensory loss, etc.) or extrinsic (associated with conditions in the environment, e.g., slippery surfaces, darkness, etc.). In community dwelling older adults, most falls (55%) are the results of an interaction between the two types of risk factors, for example, an older adult with an inferior visual field cut who fails to see a hazard on the ground, trips on the obstacle, and falls (Overstall, 2004). Spontaneous, intrinsic falls (e.g., due to head turning or legs collapsing) constitute 39% of falls. Elderly recurrent fallers are often found to have multiple intrinsic risk factors, the most critical being leg weakness, history of falls, gait deficit, balance deficit, use of assistive device, visual

deficit, arthritis, functional decline, depression, cognitive deficit, and age greater than 80 years. In a meta-analysis of the results from a multi-center study of optimal methods to reduce the risk and incidence of falls, Province et al (Province, Hadley et al., 1995) concluded that ‘balance deficits could have a more direct causal pathway to the generation of falls’ than other risk factors targeted in those trials.

Support for the strong relationship between imbalance and falls is consistently found in studies of the ability of both clinical and laboratory tests of balance to discriminate elderly ‘fallers’ from ‘non-fallers’. Clinical tests of balance typically consist of one or more balance and mobility activities such as standing with eyes closed, standing with reduced base of support, e.g., single leg stance, and reaching a maximum distance in one or more directions. Clinical tests known to have predictive validity include the Functional Reach/Multi-Directional Reach test, Berg Balance Scale, Timed Get-Up-and-Go Test, and Tinetti Performance Oriented Mobility Assessment, to name a few.

Laboratory tests of balance using forceplate measures of postural sway, that is, variables calculated from the trajectories of either center-of-pressure, COP, or center-of-mass, COM, can also discriminate “fallers’ from ‘non-fallers’. Though postural sway in quiet stance alone is a poor predictor of age category or fall status (Prieto et al, 1996), certain postural sway measures taken in conditions of heightened balance challenge, e.g., with sensory manipulation such as eye closure or increased biomechanical control demands by reducing the base of support, are discriminatory (Maki et al, 1990; Topper et al, 1993; Wallmann, 2001). Laboratory tests of balance tend to have higher sensitivity (true positives, i.e., better at classifying fallers as fallers) while clinical tests of balance tend to have higher specificity (true negatives, i.e., better at classifying non-fallers as

non-fallers). These studies of both clinical and laboratory tests of balance suggest that deficiencies in balance are strongly associated with the risk of falls in older adults.

Additional support for the relationship between imbalance and falls is offered by prior research showing that the remediation of intrinsic risk factors is a more effective approach to the prevention of falls than environmental modification (Gill, Robison et al., 1999). Specifically, comparisons of interventions to address multiple intrinsic risk factors find that various balance and strength exercises, but not other forms of exercise (e.g., walking, aerobics, conditioning, or flexibility), lead to significant reductions in the risk of falls (Province, Hadley, Hornbrook, Lipsitz, Mulrow, Ory, Sattin, Tinetti, and Wolf, 1995; Whipple, 1997). Strengthening exercises by themselves do not appear to improve balance, and vice versa (Wolfson, Whipple et al., 1993), however, each contribute to fall risk reduction by addressing different independent intrinsic risk factors. Study outcomes indicating that participation in balance exercise improves balance, and improved balance reduces the risk and incidence of falls (Gillespie LD, Gillespie WJ et al., 2003), provide further evidence for the relationship between imbalance and falls.

What causes imbalance in fall-prone older adults?

Just as falls result from multidimensional risk factors, imbalance in older adults results from multiple impairments in the postural control system. This view, now widely accepted, was first formally detailed by Horak and colleagues in their seminal 1989 paper “Postural dyscontrol in the elderly”. Horak et al were the first to apply a systems model of postural control to the specific problem of imbalance and falls older adults. Within this framework, postural control is broadly divided into sensory and motor components, each

of which has peripheral and central nervous system components. Peripheral sensory components are responsible for the detection and transmission of information from the environment to the central nervous system, and include the somatosensory, visual, and vestibular systems. Central sensory components receive, transduce and fuse this information to form the perception of body position and motion in space (multi-sensory integration), and also adapt this integration process when changes in the external environment or the task-goal alter the availability or usefulness of incoming sensory information (multi-sensory re-weighting). Central motor components include the coordination, latency, and scaling of automatic postural reactions (e.g., ankle, hip and stepping strategies), anticipatory postural responses to expected postural changes, and the initiation and control of voluntary postural motion. Peripheral motor components comprise the effector system and include the neuromuscular and musculoskeletal system of peripheral motor nerves, the muscles they enervate, and the joints controlled by these muscles. Naturally the sensory and motor systems do not function separately; sensori-motor integration processes are thought to map perception onto action.

The central sensory and motor components of postural control have been less well studied than their peripheral counterparts due to limitations in methods of investigating central nervous system (CNS) processes. Newer methods of study including quantitative modeling now advance our understanding potential CNS processing mechanisms. We use such methods here to study the central sensory and motor components of postural control, termed ‘estimation’ and ‘control’, respectively. For example, Oie, Kiemal and Jeka (2002) were able to demonstrate that changes in the relationship of postural sway to visual and touch stimuli (as represented by gain values) were due to changes in sensory

weights on the estimation side of the model, not changes in control parameters such as stiffness and damping on the control side of the model. The ability to distinguish between the postural responses to vision versus touch, and between changes in estimation versus control parameters, are far less possible with traditional measures of overall sway that represent the total outcome of both sensory and motor [estimation and control] systems.

Evolution of the systems model from earlier theoretical motor control frameworks

The systems model of postural control described above is itself evolved from prior models of postural control based on the prevailing motor control theories of the day (Shumway-Cook and Woollacott, 2001). Postural control research questions, as well as clinical assessment and intervention methods for persons with deficient postural control, have been heavily influenced primarily by four previous theoretical frameworks of motor control theory. Predominant for the first half of the 19th century was Sherrington's reflex theory, which held that reflexes were the foundation for all complex behaviors. Reflexes were comprised of a sensory stimulus from the environment, a pathway of transmission, and a response. Multiple reflexes could be triggered in sequence or 'chained', to produce more elaborate and complex motor behavior. Indeed the response to one stimulus could and did serve as the stimulus for the next reflex, and so on. Subsequently this idea that movement was initiated and controlled by peripheral sensory stimuli was modified by the work of Jackson and Magnus. They placed far greater importance on the central nervous system (CNS). The CNS was thought to use higher, middle and lower centers to exert control over the motor system in a strict, vertical, hierarchical, top-down manner, hence the 'hierarchical control theory'. One of the functions of the higher centers was to inhibit

or control the next lower center, and reflexes controlled by one of the lower centers would not be present (in adults) unless damage to one of the higher centers had released the lower center from control. While this reflex/hierarchical control theory was in vogue, efforts to understand postural control and to use this understanding in the evaluation and treatment of imbalance were thus focused on reflexes such as the vestibular-ocular and vestibulo-spinal reflexes (VOR and VSR, respectively), as well as righting and equilibrium reactions. Intervention goals emphasized the inhibition of ‘abnormal’ reflexes and the facilitation of ‘normal’ ones.

The systems model of postural control incorporates but is not limited to these reflexive peripheral and hierarchical central nervous system components. For example, it is understood that automatic postural reactions are primarily triggered by peripheral somatosensory information, and that they are controlled by ‘long-loop responses’ that involve lower brain centers but not cortical processing. However, in contrast to the reflex/hierarchical model, the systems model also incorporates the intrinsic biomechanical effector system and cognitive process components, the capacity for adaptation and learning, and the concept of distributed control. Perhaps the most striking distinction between the reflex/hierarchical and systems models is the emphasis in the latter on the influence of the task and the environment on postural control. In the evolution from the traditionally predominant reflex/hierarchical model to the current systems model we may see the influence of other important approaches to motor control.

From Bernstein’s original systems theory we obtain the increased focus on the role of the physical biomechanical system and the notion that motor control is not just “top down” but distributed and flexible. He posited that movement was highly influenced

by the initial conditions of the body to be moved, e.g. whether a limb was fixed on a surface or free in space, and that internal and external forces acting on the limb could alter motion differentially and had to be taken into account. Famously, he articulated the ‘degrees of freedom’ problem – just how can the CNS possibly individually control every muscle at every joint in every position at every speed at each point in time? His proposed solution was that the primary function of control processes is to reduce these degrees of freedom by using muscle synergies, groups of muscles constrained to act together in a coordinated and sequential fashion. Of particular interest here, Bernstein noted that there were ‘postural synergies’. His theory has strongly influenced postural control research and can be seen extensively in the investigations of these postural synergies by many of today’s motor control researchers (cf, Nashner, Horak, S-C, Woollacott, Allum, etc.).

Two additional theories have helped to strongly shape the systems model of postural control described by Horak et al. First, from perceptual psychology came Gibson’s ecological theory. He (also famously) posited that all action occurred within an environmental context integral to the action itself, that is, for an individual to achieve desired goals in the world, successful interaction with the environment was prerequisite. Information relevant to the desired objective had to be gleaned from the environment and was then used to control movement. Likewise, our movements in and through the environment allowed active exploration of the environment to gather more information. Gibson’s major contribution was his strong emphasis on the bi-directional inter-relationship between perception and action. Results from postural control studies in which the environment is systematically manipulated, and of individuals who have difficulty detecting or perceiving sensory information in the environment, support his

premise that the control of action is heavily influenced by environmental context. Secondly, a task-oriented approach to motor control was developed (Shumway-Cook & Woollacott, 2001). In this view, movement occurs to accomplish desired goals. The role of the CNS in motor control is to solve motor problems to permit successful implementation of goal-directed actions. Several postural control researchers have proposed that one critical problem to be solved is how to keep the COG over the base-of-support to prevent balance loss; others define the primary problem as control of head position in space to optimize visual and vestibular functions (Horak, Nashner et al., 1990). The sensori-motor processing component of the systems model provides a mechanism for the perception-action relationship developed by Gibson. The systems approach also takes the broader view that these intrinsic postural control system components are affected by and responsive to the environmental context, and that they operate together in order to accomplish various functional tasks in real life that pose problems for the system to solve.

The systems approach described by Horak and others (cf, Nashner, Shumway-Cook, Woollacott) has evolved from these earlier theoretical frameworks and incorporates concepts from several of them. In the following section we will elaborate on this systems model of postural control, discuss the normal function of the different components, and describe how these components appear to change with age and subtle pathology.

What is balance? A systems approach.

Within this multidimensional systems model, postural control serves two main functions (Horak and Macpherson, 1996). One primary function is orientation, including both orientation of the body segments to each other (“posture”), e.g., alignment of the trunk over the legs, the head over the trunk, etc., as well as orientation of the body whole in space, e.g., its’ relationship to the support surface and gravitational vertical, and to hazards and affordances in the environment. The second primary function is to establish and maintain stability while allowing volitional movement, that is, to keep the center of gravity over the current or expected base of support to prevent loss of balance (“control”). How the postural control system solves these biomechanical challenges is heavily influenced by the environment in which the individual is standing and moving, and by the task goals the individual wishes to accomplish.

Horak et al. have proposed that to solve the dual challenges of orientation and stability, the postural control system uses sensory, motor, and sensori-motor strategies. Sensory strategies are used to organize, integrate and re-weight multiple sensory inputs, form internal representations of the self in space to ‘map’ perception to action, and to anticipate and adapt to changes in the sensory environment. Motor strategies are used to organize and control the position and motion of the body in space, including reactive strategies for unexpected disturbances of stability and pro-active strategies for expected disturbances, either external or self-generated by volitional movements. Sensori-motor strategies are used to coordinate the two processes.

Peripheral sensory components

Successful interaction with the real world around us demands a continuously updated awareness of the surrounding environment. The role of the peripheral visual, vestibular and somatosensory systems in postural control is to detect information from the environment and transmit it to the CNS for subsequent processing. Peripheral sensory information is used by lower levels of the CNS to trigger automatic postural reactions, and by higher levels of the CNS to develop internal representations of the self in space for the generation of anticipatory postural responses and voluntary movements (Tang and Woollacott, 2004). No single sense can fully inform the CNS; while some information from each separate sense is unique, there is partial redundancy of information from the three senses. For example, sensory inputs from the visual and vestibular otolith systems provide information about low frequency motion, while inputs from the somatosensory and vestibular semi-circular canal systems provide information about high frequency motion. This redundancy serves a necessary purpose, to resolve ambiguities that arise from a single sensor. For example, the vestibular system alone cannot distinguish whether forward head motion in space is caused by flexion at the neck or flexion at the hips; information from somatosensory receptors in the neck and trunk can be used to make this distinction. Peripheral sensory loss affects the redundancy inherent in and important to the postural control system.

The somatosensory system is the most critical sense for upright postural control. Blind persons can stand and walk, individuals with [chronic] bilateral vestibular loss can function well except in quite compromised environments, but major bilateral loss of somatosensation will preclude any sort of normal posture or gait. Somatosensation

includes the senses of vibration, joint proprioception, touch and pressure, and muscle length and tension. Motion of the COG over the feet will produce changes in the sense of touch and pressure on the plantar foot surface, the length and tension in the gastrocnemius muscle, and the deformation of the ankle joint capsule. This somatosensory information appears critical for the detection of subtle postural sway, the triggering of automatic postural reactions, and adaptation to repeated visual postural perturbations (Colledge, Cantley et al., 1994; Inglis, Horak et al., 1994). The removal of somatosensation leads to significantly increased sway, while the addition of somatosensory information, whether at the feet or fingertip, and whether active or passive, improves sway control (Hay, Bard et al., 1996; Jeka and Lackner, 1994; Jeka, 1997; Jeka, Schoner et al., 1997)

Somatosensory loss is common with increasing age, even in otherwise healthy older adults. In the healthy 'young old', ages 65-74, the incidence of vibration loss is 12%, proprioceptive loss is 21%. In the healthy 'oldest old', those over age 84, 68% have vibration loss, and 44% have proprioceptive loss (Kaye, Oken et al., 1994). Motion detection and position replication at the knee are more impaired at slow versus fast speeds. Diminished vibration and proprioception are both associated with increased sway, decreased stability (particularly in the medio-lateral direction), and falls (Duncan, Chandler et al., 1993; Hughes, Duncan et al., 1996; Lord and Ward, 1994). Tactile sensitivity also declines with age. Somatosensory loss is more extensive in fallers than non-fallers; significant loss is frequently associated with pathology.

Visual inputs also contribute to postural control. Foveal or focal vision is used for navigation in the environment, including the detection of hazards and affordances

therein (Tang and Woollacott MH, 2004). It is also uniquely critical for feedforward postural control, that is, for prediction of upcoming environmental conditions and potential extrinsic perturbations. Ambient or peripheral vision is used for the detection of our own intrinsic motion as well as movement in the environment. Some studies show that a reduction in visual information leads to increased sway, but others find no such association in healthy adults on stable surfaces.

Visual changes with age are well-documented. Visual acuity declines and is worse in fallers than non-fallers (Lord and Ward, 1994). Loss of acuity, however, may be less related to imbalance and falls than other common visual impairments. Of greater impact are visual field loss, diminished depth perception and contrast sensitivity, decreased sensitivity to self-motion and object motion in the environment, and reduced ability to perceive extrinsic horizontal and vertical cues (worse in fallers than non-fallers) (Overstall, 2004;Tang and Woollacott MH, 2004;Whipple R, 1989). In studies that used a moving room paradigm (discussed more fully in the next chapter), older adults have been found to be more sensitive to visual motion, particularly in conjunction with somatosensory loss (Borger, Whitney et al., 1999;Colledge, Cantley, Peaston, Brash, Lewis, and Wilson, 1994;Lord and Ward, 1994;Sundermier, Woollacott et al., 1996).

Vestibular information is used to sense the position and motion of the head (Horak, Shupert et al., 1994;Horak, Buchanan et al., 2002). It functions to maintain gaze stabilization during head movement, and, in conjunction with neck somatosensation, to monitor and control the position of the head in space. Some researchers have argued that the vestibular system serves as an absolute orientation reference to resolve sensory conflicts, since it is the only sense attuned to an unchanging sensory input, gravity

(Nashner, Black et al., 1982; Nashner, Shupert et al., 1989). Other scientists have concluded that the somatosensory system serves that purpose {Bronstein, 1999; Anastasopoulos & Bronstein, 1999). With sudden unexpected changes in head position, vestibular inputs can trigger automatic postural reactions. Sudden loss of vestibular inputs is initially destabilizing, but in most individuals who do not restrict their motion, rapid adaptation occurs and thereafter these individuals are able to develop adequate postural control except in circumstances where neither visual or somatosensory inputs are available or accurate.

Age-related changes in the vestibular system may be more pronounced physiologically than functionally. A significant 20% to 40% loss of vestibular hairs cells, neurons and nuclei has been documented in healthy older adults, however, age effects on vestibular function tests appear minor (Peterka and Black, 1990b; Peterka, Black et al., 1990). Many older adults report dizziness (a common symptom of vestibular involvement), but only a relatively small subset (20%) of these were found to have diagnosable vestibular pathology (Belal & Glorig (1986) as cited in Horak, 1990 (Horak FB, Mirka A et al., 1990). Wolfson has stated that there are “no data to suggest [vestibular loss] is a major factor” in age-related balance changes (Wolfson L, 1997) p. 81. Others disagree based on findings that vestibular function tests are correlated with balance performance in healthy older adults on compliant surfaces with eyes closed, and that individuals with known peripheral vestibular deficits have difficulty suppressing unreliable visual or somatosensory inputs (Lord and Ward, 1994). Further, in contrast to the Belal study, Mirka found a high incidence (60/65) of vestibular ocular or postural control deficits in older adults with dizziness (Mirka et al, 1988, as cited in Horak, 1990

(Horak FB, Mirka A, and Shupert CL, 1990)). Horak has suggested that age may affect oculo-motor versus postural control functions differently, citing a report by Peterka (1987) in which the vestibular-ocular reflex (VOR) of subjects who fell in SOT conditions 5 and 6 [when both vision and somatosensory information were absent or unreliable] was not different from those who did not fall (Horak FB, Mirka A, and Shupert CL, 1990).

Multi-sensory deficits: While a reduction or loss of any one of these three senses will certainly affect postural control, there is a consensus that it is the presence of multiple sensory impairments that most negatively impacts older adults. Multi-sensory deficits are common in the elderly. Their combined effect negatively impacts sensory redundancy and thus the ability to resolve ambiguities and sensory conflicts (Horak FB, Mirka A, and Shupert CL, 1990).

Central sensory components

Upon arrival in the CNS, these three separate senses are merged or fused to form a single perception and internal representation of body position and motion in space; this process is termed multi-sensory integration. Accurate perception of body position and motion is a prerequisite for normal postural control since motor commands to the musculature to counteract gravity and generate volitional movement depend on the correct estimation of current and future body state. The integration process is straightforward when the three separate senses are equally valid and in agreement, that is, when the information from each modality is reliable and consistent with the information from the other two. Additional processing is required however, if the relative reliability

of the senses is changeable, or if the senses conflict. In situations where the reliability of one sense fluctuates relative to another, e.g., sudden darkness makes visual information less useful for stability, the CNS must adapt by placing a lower priority on vision and a higher priority on the available and accurate somatosensory and vestibular inputs.

Likewise, in situations of sensory conflict, the CNS must first recognize the discrepancy, down-weight (suppress) the inaccurate input, and up-weight the accurate inputs. This process is multi-sensory re-weighting, and we will discuss it in more detail in the next chapter (Horak FB, Mirka A, and Shupert CL, 1990; Nashner LM, 1997; Tang and Woollacott MH, 2004; Wolfson L, 1997).

Central sensory organization and perception of stability limits are also diminished in the aged. Numerous experiments in which visual and somatosensory inputs are systematically manipulated suggest that failure to compensate for sensory loss and deficits in sensory conflict resolution increase with age and are associated with imbalance and falls in the elderly (Alexander, 1994; Whipple, Wolfson et al., 1993; Wolfson, Whipple et al., 1992; Woollacott, 1993). When presented with conflicting sensory input, young subjects sway significantly more, but do not fall, while older subjects often lose balance, especially on the first trial of a novel condition (Judge, King et al., 1995; Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992; Woollacott, Shumwaycook et al., 1986). Loss of balance during these trials is more prevalent in fallers compared to healthy older adults (Anacker and Di Fabio, 1992). Healthy older adults demonstrate instability only in conditions where two senses are manipulated simultaneously, while fall-prone elders also become unstable when any one sense is altered (Judge, King, Whipple, Clive, and Wolfson, 1995; Manchester,

Woollacott et al., 1989). Several studies have also reported an abnormal reliance on vision in the elderly, with inaccurate or suddenly altered visual inputs having a more destabilizing effect than no visual input (Horak, Shupert, and Mirka, 1989; Simoneau, Teasdale et al., 1999; Sundermier, Woollacott, Jensen, and Moore, 1996; Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992). The ability to select and prioritize sensory inputs adaptively appears to be slowed in almost all older adults, and significantly reduced or absent in elderly fallers (Camicioli, Panzer et al., 1997).

Inaccurate perception of the limits of stability has been documented in older adults; the limits are perceived to be smaller than they actually are, and the older adult will not lean very far away from midline for fear of exceeding the (perceived) stability boundary (Blaszczyk et al, 1993). While older adults may not be able to lean as far forward as young adults, elderly fallers lean significantly less than age matched non-fallers (Duncan, Studenski et al., 1992). It is unclear at this time whether this finding reflects an inability (due to perceptual problems) and/or an unwillingness (due to fear) to move the COG away from the midline position.

Central motor organization

Subsequent to sensory integration, neural commands for volitional movements, anticipatory postural adjustments and automatic postural reactions must then be generated in the central nervous system. At this stage in the postural control process, the CNS ‘knows’ where the body is located in space, and how it is moving through space. Changes in the position and trajectory of the COM may be desirable, however, when the COM has been unexpectedly disturbed, or because the CNS expects an upcoming disturbance to the

COM from external forces or planned, self-initiated movements. In the case of large or fast unexpected perturbations, automatic postural reactions are used to rapidly re-establish the stable relationship of the COM over the base of support, or to create a new base of support under the ‘run away’ COM. When perturbations are expected in advance from extrinsic or intrinsic sources, neuromuscular preparation for such predicted disturbances also occurs in advance; these are anticipatory postural adjustments. Lastly, when volitional, goal-directed movements of the body in space are planned, e.g., shifting the COG to move dishes from the dishwasher on the lower left to the cupboard on the upper right, the CNS must initiate and control the motion of the COM through space.

Automatic postural reactions have traditionally been tested by applying an unexpected mechanical perturbation to the upper or lower trunk segment, or more commonly to the support surface (Horak and Nashner, 1986; Horak, Henry et al., 1997). Typically, but not uniformly, the perturbation is given in the anterior/posterior (A/P) direction. Depending on the relative amplitude and velocity of the perturbation, and given the desire of the individual to maintain a stable position, ‘stereotypical’ movement patterns, called ‘strategies’ may be observed.

If the support surface is broad and stable and the A/P perturbation is small and slow, such that the COG is disturbed away from its’ desired location but does not approach the stability limits, the “ankle strategy” is seen. This movement pattern is characterized by in-phase motion of the head and pelvis, and is controlled primarily by ground reaction forces produced by the ankle. If the support surface is smaller than the base of support, or unstable, and/or the A/P perturbation is somewhat larger and faster such that the COG is disturbed and approaches the stability limits, the “hip strategy” is

seen. This movement pattern is characterized by out-of-phase motions of the head and pelvis, and is controlled primarily by shear forces produced by the hip. If the individual is not especially constrained to remain in place, and/or if the amplitude and velocity of the perturbation are so large as to shift the COG past the limits of stability, a “stepping” or “reaching” strategy is observed. These are attempts by the individual to re-establish a new base of support underneath the moving COG, to prevent a fall.

Automatic postural reactions were initially thought to be distinct and inflexible in nature, but these ideas have been revisited and are evolving (Tang and Woollacott MH, 2004). Currently it is accepted that these movement patterns occur on a continuum, and that observed strategies will vary depending on the characteristics of the individual, the instructions they have received, and the parameters of the perturbations. Very recent work indicates that even in quiet standing, with no other perturbation than the steady action of gravity on the body, both ankle and hip strategies are simultaneously present as excitable coordination modes (Creath et al, 2005).

Age differences are apparent in the selection and implementation of automatic postural reactions. When constrained to remain in place, healthy older adults more often use hip strategy than young adults (Horak, Shupert, and Mirka, 1989; Manchester, Woollacott, Zederbauer-Hylton, and Marin, 1989). When permitted or obligated, both young and healthy older adults use a stepping strategy given sufficient A/P perturbation; young adults tend to take a single large step while older adults take multiple steps and often lateral steps (McIlroy & Maki, 1996). For medio-lateral (M/L) perturbations, older adults also take multiple versus single steps, produce inadvertent collisions between feet when stepping, and use more upper extremity compensatory movements (Maki et al,

2000). Whipple et al. have reported a correlation between the propensity to fall and a larger number of steps taken, or the failure to step at all, in response to a posterior perturbation of the trunk (Chandler, Duncan et al., 1990).

While the motor strategies described above are visible to the naked eye, they are associated with and produced by **muscle synergies** that are detectable by electromyographic (EMG) recordings. Ankle and hip strategies result from combined muscle actions to move the COG in the direction opposite the perturbation. If the perturbation moves the COG forward, the activated muscles will be those whose combined contraction will move the body backward, and vice versa. In healthy young persons, these synergies are activated within 160 msec. and the size of the response is scaled or matched to the size of the perturbation. Within the synergy, the timing and sequence of muscle contractions is coordinated. In muscle synergies associated with the ankle strategy, the distal leg musculature is activated first, followed by the proximal leg muscles. For the hip strategy, the proximal leg muscles are activated first, followed by the distal musculature. There is typically a 30 to 50 msec difference in contraction onset between the distal and proximal muscles. Unless the perturbation itself shifts the COG sufficiently to release weight from one of the lower extremities, the stepping strategy involves a rapid sequence of muscle contractions that first move the COG over the leg that will step, followed by a push-off from the stepping leg that moves the COG over the stance leg, and then a lift-and-place motion of the stepping leg in the direction of the impending balance loss.

Compared to young adults, older adults demonstrate slightly slower onset latencies in the distal leg muscles, decreased magnitude and longer duration of muscle

responses, dyscoordination of muscle contraction sequences and timing, and more frequent co-activation of both agonist and antagonist muscles (Horak and Nashner, 1986;Manchester, Woollacott, Zederbauer-Hylton, and Marin, 1989;Peterka and Black, 1990a;Tang, Woollacott et al., 1998;Woollacott, Shumwaycook, and Nashner, 1986). These findings suggest age-related changes in the central motor mechanisms of postural control, however, peripheral motor changes might also help to explain these results (Tang and Woollacott MH, 2004).

Sensori-motor adaptation has traditionally been tested with repeated exposure to sudden support surface rotations, which rapidly alter the ankle joint angle and ankle musculature length but produce little shift in the center of gravity. (Horak and Nashner, 1986;Horak, Diener et al., 1989;Nashner, Woollacott et al., 1979;Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992). Initial responses to this ‘false’ support surface transition in young healthy subjects are large (relative to the actual COG perturbation stimulus), but are very rapidly attenuated in three to ten trials if the same condition is repeated consecutively. In older adults, adaptation to repeated surface rotations is present but diminished. Compared to young adults, older adults are more likely to lose their balance on the first exposure, but do adapt quickly thereafter, though they may require more repetitions prior to full adaptation (Woollacott, Inglin et al., 1988;Woollacott, Shumway-Cook et al., 1986).

Anticipatory postural adjustments are, of course, made prior to observable limb or body motion, and therefore are typically evaluated with EMG. They reflect the ability of the CNS to predict and prepare for limb or body motions that are likely to disturb the position or motion of the COG. They function to help counteract these COG

perturbations and permit smooth integration of the dual postural control goals of orientation and stability during motion. Anticipatory postural adjustments occur in advance of the expected perturbation, and are specific to it (Horak et al , 1989). For example, prior to lifting a heavy object up from the floor (which would tend to pull the COG forward) the CNS may initiate a backward shift of the COG to counteract the expected forward force.

Age-related changes in anticipatory postural control are minimal under conditions where the task is predictable and the timing is self-paced. In choice reaction time tests involving standing and pushing versus pulling on a lever, however, older adults show increased latencies for both the postural muscles in the trunk and legs and the prime mover muscles in the arm (Inglin and Woollacott, 1988). The normal distal to proximal muscle contraction sequence in the legs was also disrupted. In a different reaction time task in which young and older adults were asked to lift one leg as rapidly as possible, increased onset latencies were observed in the muscles of the [postural] stance leg, but not in the [prime mover] lifted leg. This resulted in a shorter gap between the stabilizing contractions in the stance leg and the destabilizing motion of the lifted leg (Mankovskii (1980) as cited in Woollacott , 1990) (Woollacott MH, 1990).

Voluntary reaction times are also increased in older adults over 60 years of age (Wolfson L, 1997). The afferent and efferent transmission of neural inputs and outputs changes little with age, but moderate changes may be seen in the transduction of stimuli into nerve impulses and the time required to activate muscles. Sensori-motor processing, which represents the largest part of reaction time, is the most affected by advancing age (Stelmach and Worringham, 1985;Woollacott, 1993). Significant increases in sensori-

motor processing time are especially noticeable in choice and complex reaction time tasks.

Peripheral motor components

The neural commands generated in the CNS are then transmitted to the peripheral musculoskeletal system and implemented to produce movement. This effector system functions to generate forces to move and stabilize the head, trunk and limbs in space. Age-related changes in the musculoskeletal system are well-documented. Lower extremity strength is reduced by as much as 40% in healthy older adults over age 80. Lower extremity weakness is prevalent in the fall-prone elderly; nursing home residents are two to four times weaker than healthy older adults living in the community (Gehlsen and Whaley, 1990; Whipple, Wolfson, and Amerman, 1987). Muscle mass is also decreased, though not to the same extent as muscle strength. Reductions in the speed of contraction, peak torque, and power occur; all are crucial for the rapid recovery of equilibrium following loss of balance. Joint flexibility and range-of-motion are also decreased in older adults; fall-prone elders are noted to have flexed posture and a loss of extension excursion compared to healthy older adults (Gehlsen and Whaley, 1990). Finally, falls are more likely to occur with fatigue; low endurance is an additional risk factor for falls.

The cognition component

The influence of cognitive load on postural control has been studied using the traditional dual-task paradigm. Normally, balance is automatic and its' control does not

require conscious attention unless the balance task is especially challenging. In older adults with diminished intrinsic balance control, even ‘usual’ daily activities requiring balance, e.g., donning trousers while standing up, can demand focused attention. Older adults with imbalance demonstrate a reduced ability to appropriately allocate attention away from secondary tasks to a balance task causing instability (Maylor and Wing, 1996; Shumway-Cook, Woollacott et al., 1997). Shumway-Cook et al (Shumway-Cook and Woollacott, 2000) found that for young adults, performance on the SOT was not compromised by the addition of a secondary task. Healthy older adults’ performance declined only in the two more challenging conditions, and the performance of fall-prone elders was reduced in all six conditions. The ability to generate successful automatic postural responses is also compromised in older adults when their attention is diverted to a secondary task (Rankin, Woollacott et al., 2000).

In summary, impairments in any component of the postural control system (e.g., sensory loss, perceptual deficits, slowed central processing, muscle weakness) can lead to one or more balance deficits. There is an accepted consensus that healthy elderly individuals have fewer, less severely involved components, compared to the unstable elderly or those experiencing falls. Certain researchers have focused on the relative degree of risk posed by some impairments versus others, yet all appear to concur that a greater number of impairments (and their interactive effects) directly increase the risk of falls. The concept of “functional reserve” as a critical threshold for functional loss and falls has been proposed (Duncan, Chandler, et al, 1993). In other words, one-plus-one equals more than two: the presence of multiple impairments (versus which particular impairments) may be the critical factor.

Measures of postural control

Postural sway is the [usually] unconscious, constant, slow, small motion of the body about the vertical midline. It occurs naturally as the body has mass which is acted upon by gravitational forces that pull the upright body away from the midline, and also has muscles which generate forces to hold the body upright and push against support surfaces to counteract the effects of gravity on the body. Postural sway is the result of these opposing extrinsic and intrinsic forces. Traditionally, measures of postural control are measures of postural sway itself, or of the muscles that contract to actively produce or reactively counteract it.

Prior to technological advances, direct measures of postural sway were made mechanically with tracing techniques using brushes or pencils attached to the head, shoulders or trunk (Sheldon, 1963). Such methods were used until relatively recently. As technology was developed and improved, more sensitive and accurate measures of sway became possible. The use of single or dual forceplates is now widespread in research laboratories, and commercial forceplate systems are also gradually gaining limited acceptance in clinical settings. Forceplates detect the ground reaction forces and moments in three movement planes and permit the calculation of the center-of-pressure (COP), a representation of body motion in space detected at the interface of the feet and the support surface. From the COP, assuming a known base-of-support and using subject height, the center-of-mass (COM) may also be calculated. In quiet standing the correlation between the COP and the COM is quite high, but this relationship becomes more divergent as body motion increases. The position and motion of COM may also be measured separately from the COP using motion analysis. One or more markers may be

placed on the body, and the motion of the markers captured on film or onto a computer drive. When a single marker is used it is usually placed at the lower lumbar/upper sacral vertebrae, to approximate the location of the COM in a quietly standing person. More accurate approximations of the location of the COM may be obtained by using multiple markers to identify linked body segments and calculating the position of the COM using known biomechanical formulae, e.g., Winter et al, Chiari et al, etc.

From the trajectory of either the COP or COM, variables to quantitatively describe postural sway behavior may be calculated. In the time domain these typically include sway area, sway area variability, sway envelope (perimeter length), path length (total), sway amplitude (either away from a midline point, from the mean COP/COM position, or peak-to-peak range), and sway velocity. In the frequency domain commonly used variables include power spectral density, power frequency, zero-crossing frequency, and frequency dispersion. To obtain a quantitative measure of the relative ‘coupling’ of the intrinsic central sensory system and an extrinsic source of sensory information, e.g., vision and touch, we have used a transfer function from which we calculate the ratio of the amplitude of body sway at the driving frequency of the sensory stimulus to the amplitude of the sensory stimulus itself; this ratio is the gain. A gain of one reflects that the amplitude of the postural response at the sensory drive frequency is equivalent to the sensory drive amplitude. We also obtain a quantitative measure of the temporal relationship of the body sway to the sensory stimulus, phase.

Most prior studies of multi-sensory re-weighting in the elderly have used as their outcome measures traditional time domain measures of postural sway. We have been critical of this approach, as such overall measures of postural sway reflect all

contributions to postural sway - sensory, motor, and biomechanical – and cannot discriminate sensory contributions to sway from other influential components of the postural control system. For example, many studies have used laboratory or commercial versions of the Sensory Organization Test® to investigate multi-sensory re-weighting. This test involves a moving, sway-referenced support surface that functions to render somatosensory inputs inaccurate. Other studies have used a compliant foam support surface for the same purpose. Both the sway-referenced forceplate and the compliant foam introduce not only sensory conflict but also instability and biomechanical challenges that require control responses. These challenges may be essentially inconsequential to an individual with normal reaction time and lower extremity range of motion and strength, but may lead to increased sway in persons with slow reactions times and compromised peripheral motor systems. Conclusions from prior studies that an increase in overall sway following sensory input manipulations necessarily reflects primarily a sensory deficit must thus be scrutinized carefully, especially if the subjects studied had impairments in motor and biomechanical components of their postural control systems, as fall-prone older adults often do. The use of gain and phase variables derived from frequency analysis of postural sway at the sensory drive frequencies permits a more precise and discriminatory ‘capture’ of the central sensory processing component separate from the other postural control system components that are also known to influence postural sway.

How well do different measures of postural control distinguish between young and old age groups, and between healthy and fall-prone older adults? In general, static measures of postural sway under low-challenge conditions are poor discriminators of age

and predictors of fall status (Alexander, 1994; Alexander, 1996; Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992). Prieto and colleagues found that COP sway velocity was the only ‘quiet stance’ measure that was significantly different in healthy young versus older adults, and this measure also discriminates fallers versus non-fallers in institutional settings (Fergie, Gryfe et al., 1982; Prieto, Myklebust et al., 1996).

The addition of sensory or biomechanical challenges improves age discrimination and fall-risk prediction (Topper, Maki et al., 1993; Whipple, Wolfson, Derby, Singh, and Tobin, 1993). Anterior/posterior sway area and velocity are increased in healthy older versus young adults when vision is absent or the surface is in motion or sway-referenced (Baloh, Fife et al., 1994; Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992). Medio-lateral sway measures in quiet stance appear better able to discriminate fallers versus non-fallers (Lord and Ward, 1994; Maki, Holliday et al., 1990).

Postural sway in the SOT conditions where only one sense is manipulated is similar in healthy young and older adults, but fallers have greater difficulty than their healthy peers in these conditions. In all age groups, postural sway is greater in SOT conditions five and six, where both vision and somatosensation are concurrently manipulated. Young adults very rarely lose balance in this condition. Thirty to fifty percent of healthy older adults may lose balance on the first trial of these conditions, but rapidly adapt and do not become unstable on subsequent trials. Fall-prone, but not healthy, older adults also lose balance on subsequent trials of these same conditions, reflecting potential inability to reweight, to adapt, or to use vestibular inputs.

The addition of biomechanical challenge by reducing the size of the base of support also helps to discriminate between age groups, and predict fall-risk. Young adults

have no difficulty in maintaining single leg stance for 30 seconds, but mean single-leg stance time with eyes open is only 22 seconds for adults in their 70's and 14 seconds for adults in their 80's. Fall-prone older adults are typically unable to hold this position for 5 to 10 seconds. Sway path length in single leg stance is significantly greater in older versus younger adults (Bohannon, Larkin et al., 1984; Bohannon, 1994).

Age differences are also apparent in perturbation tests of reactive postural control. Healthy older adults have greater sway area, higher sway velocity, and more frequent loss of balance than do young adults after large fast perturbations (Baloh, Fife, Zwerling, Socotch, Jacobson, Bell, and Beykirch, 1994; Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992). Older adults use more hip strategy than young adults when instructed not to step. When reactions are unconstrained, older adults, and especially fallers, tend to react with multiple, very rapid small steps versus the single large step taken by young adults. Fall-prone older adults, but not their healthy peers, sometimes fail to step at all and exhibit a "timber" response (Chandler, Duncan, and Studenski, 1990). These differences are exacerbated as perturbation size and speed increases. Stelmach and colleagues have found that healthy older adults who are exposed to fast perturbations that are superimposed on small, slow surface oscillations or during voluntary movements do not react as quickly or adapt to repeated exposure as well as young adults do (Stelmach, Phillips et al., 1989; Stelmach, Teasdale et al., 1989). These results suggest that lower-level, 'long loop' postural reactions are less affected by age than higher-level, sensori-motor integration mechanisms.

EMG recordings of muscle activity during perturbations show that onset latencies are not dissimilar in young versus older adults with the exception of the tibialis anterior.

Older adults have less coordinated timing of the muscle contraction sequences, more co-activation of antagonist muscles, and reversal of the typical distal to proximal sequence which may be related to the use of hip versus ankle strategy (Woollacott, Shumwaycook, and Nashner, 1986;Woollacott, von Hosten et al., 1988).

In adaptation tests with support surface toes up/down rotation, healthy older adults are significantly less stable only on the first trial, and are not different than young adults by the third trial. As with surface translations, onset latencies are similar in the two age groups. Older adults have increased antagonist activity and may display abnormal sequencing of muscle contractions (Manchester, Woollacott, Zederbauer-Hylton, and Marin, 1989;Wolfson, Whipple, Derby, Amerman, Murphy, Tobin, and Nashner, 1992;Woollacott, Shumwaycook, and Nashner, 1986).

Volitional postural motion through space diminishes with age and is more restricted in fall-prone elders. The ability to shift the COM is reduced by about 20% in healthy older adults in their 60's, and declines by 16% per decade thereafter (King, Judge et al., 1994). Maximum COM excursion is more limited in fall-prone than healthy older adults and failure to shift the COM beyond certain distance thresholds is predictive of fall-risk (Duncan, Studenski, Chandler, and Prescott, 1992;Wallmann, 2001).

Chapter 2:

Multi-sensory Re-weighting of Vision and Touch is Intact in Healthy and Fall-Prone Older Adults

1. Introduction

The ability to select and reweight alternative orientation references adaptively is considered one of the most critical factors for postural control in clinical and elderly populations (Horak, Shupert et al., 1989; Horak, Shupert et al., 1994). Separate from and in addition to the peripheral sensory loss commonly found in older adults, central sensory selection and re-weighting processes are thought to degrade with increasing age, and are hypothesized to be particularly deficient in fall-prone versus healthy older adults (Horak, Shupert, and Mirka, 1989; Teasdale, Stelmach et al., 1991b; Teasdale, Stelmach et al., 1991a; Teasdale, Bard et al., 1993; Alexander, 1994; Woollacott, 2000).

For example, Teasdale et al (1991b) found that both young subjects and older subjects, screened to insure intact peripheral sensation, had increased sway dispersion with the sudden removal of visual information. This is consistent with the idea that reduced peripheral sensory information results in a poorer central estimate of body dynamics (i.e., position and velocity) and thus less precise control of sway. Likewise, adding sensory information should improve estimation of self-motion and lead to better sway control. Teasdale's results followed this pattern with the young subjects, whose sway dispersion decreased when vision was suddenly added. However, for the older subjects, sudden addition of visual information led to increased sway dispersion.

Considering that peripheral sensation was intact in both groups, these results point to deficient central mechanisms in the older group. Teasdale et al. (1991b) concluded that compared to younger adults, older adults have “poorer central integrative mechanisms responsible for reconfiguring the postural set” [page 695].

Age-related changes in central sensory integration and re-weighting have been studied by comparing healthy young adults to healthy older adults. These studies indicate that healthy older adults are similar to healthy young adults in conditions where only a single sense is altered, but are less stable in conditions where two senses are manipulated simultaneously (Whipple, Wolfson et al., 1993;Shepard, Schultz et al., 1993;Baloh, Corona et al., 1998;Baloh, Fife et al., 1994). Many healthy older adults lose their balance on the first exposure to conditions where both visual and somatosensory inputs are altered, but show far fewer losses of balance on subsequent trials of the same conditions, indicating rapid adaptation to the altered environmental demands (Woollacott, Shumway-Cook et al., 1986;Horak, Shupert, and Mirka, 1989;Wolfson, Whipple et al., 1992).

The contribution of central sensory processing deficits to imbalance and falls in the elderly has been explored via comparisons between healthy and fall-prone older adults. Fall-prone older adults demonstrate greater instability in conditions where only one sensory input is altered compared to their healthy cohorts (Anacker and Di Fabio, 1992;Baloh, Spain et al., 1995;Shumway-Cook and Woollacott, 2000). They typically fail to adapt to altered sensory conditions, often losing balance repeatedly despite continued exposure (Horak, Shupert, and Mirka, 1989;Masdeu, Wolfson et al., 1989). Fall-prone elders are also hypothesized to be more visually-dependent, failing to use

reliable somatosensory cues in environments where visual inputs are unstable (Sundermier, Woollacott et al., 1996;Simoneau, Teasdale et al., 1999).

Are central sensory re-weighting deficits responsible at least in part for the postural control problems seen in healthy and fall-prone older adults? Previous researchers interpreted their results as evidence of central re-weighting deficits [versus peripheral sensory degradation]. We describe below two methodological limitations that leave such an interpretation open for further question.

First, several earlier studies of multi-sensory integration in older adults have either included subjects with detectable peripheral sensory deficits, or not described measures used to prevent their inclusion, which if present may have confounded their results (Teasdale, Stelmach, and Breunig, 1991a;Sundermier, Woollacott, Jensen, and Moore, 1996). Age-related peripheral sensory loss is well-documented and is hypothesized by some to be the primary reason why older adults lose their orientation sense (Woollacott, Shumway-Cook, and Nashner, 1986;Masdeu, Wolfson, Lantos, Tobin, Grober, Whipple, and Amerman, 1989;Woollacott, 1993). Naturally, central sensory integration processes cannot successfully produce accurate estimations of spatial position if incoming information from the periphery is inadequate. It follows that when changes in postural control are found in a group of older adults that includes subjects with peripheral sensory loss, these changes cannot be definitively attributed to central sensory mechanisms alone.

A second limitation in previous studies of multi-sensory integration in older adults is that these studies have used overall postural sway measures (e.g. mean sway amplitude of the center-of-mass or center-of-pressure) as the outcome measure indicating

re-weighting. Poorer performance (larger sway amplitude) under altered sensory conditions has been widely interpreted as evidence of sensory re-weighting deficits (Woollacott, Shumway-Cook, and Nashner, 1986). However, we argue that mean sway amplitude is a relatively gross ‘outcome’ measure of sway performance which cannot distinguish between several different underlying mechanisms, any one of which could affect postural sway.

For example, existing models that represent the role of the three primary sensory systems for balance (visual, vestibular and somatosensory) can reproduce the experimental observation that adding sensory information reduces the total sway variance of quiet stance (van der Kooij, Jacobs et al., 1999; van der Kooij, Jacobs et al., 2001; Kiemel, Oie et al., 2002; Peterka, 2002; Mergner, Maurer et al., 2003). However, since virtually any proposed mechanism for using sensory information would have this property, it does not help us identify the mechanism actually used by the postural control system. Hence, a reduction or increase in mean sway amplitude due to the addition or withdrawal of sensory information is not a sufficiently specific finding. If modeling is to be used to understand more specifically the multiple, individual mechanisms underlying the estimation and control of posture, additional properties/constraints are necessary. Larger mean sway amplitude of elderly vs. young subjects could be due to a number of underlying changes, including non-sensory deficits such as weakness, slow reaction time, etc. and thus cannot be definitively attributed solely to multi-sensory re-weighting (Horak, Shupert, and Mirka, 1989; Whipple R, 1989; Anacker and Di Fabio, 1992; Colledge, Cantley et al., 1994).

The purpose of the current study was to examine the assumption of impaired multi-sensory re-weighting in healthy and fall-prone older adults using a light touch/vision experimental design developed by Jeka and colleagues (Oie, Kiemel et al., 2002). A crucial aspect of the design is to present stimuli from different modalities at different frequencies so that the response to each stimulus (e.g., gain and phase) can be quantified separately, thus revealing their inherent interdependence. The design also permits a distinction between intra-sensory re-weighting, in which the response to one modality changes when that modality changes, versus inter-sensory re-weighting, in which the response to one modality (e.g., vision) changes when a second modality (e.g., touch) changes and the first modality does *not* change. Here we investigated (1) age-related changes in multi-sensory re-weighting by comparing healthy young and healthy older adults, and (2) the possibility that multi-sensory re-weighting deficits contribute to imbalance and falls in the elderly by comparing healthy and fall-prone older adults.

2. Method

2.1 Subjects

Three groups of subjects participated in this experiment. Healthy young adults ($n = 10$) were students from the University of Maryland who ranged in age from 19 to 28 years and had no known musculoskeletal or neurological disorders that might have affected their ability to maintain balance. Data from these healthy young subjects was collected and reported previously (Oie, Kiemel, and Jeka, 2002) and is used herein solely for comparison. Healthy older adults ($n = 15$) with a mean age of 79 ± 3 years, and fall-

prone older adults ($n = 28$) with a mean age of 83 ± 4 years, were recruited from a large congregate retirement community, University of Maryland alumni, and the local community.

All elderly volunteers completed an eligibility questionnaire and were excluded from the study if they had any of the following: known visual conditions or impairments affecting daily function (e.g. reading, driving, etc.); dizziness, vertigo or known vestibular conditions or impairments; lower extremity numbness or conditions leading to somatosensory loss; neurological disorders; cognitive decline; low endurance or inability to independently assume and maintain the stance position required for experimental testing. Older subjects with chronic and stable orthopedic conditions, e.g. osteoarthritis of the knee, were included while those with recent musculoskeletal changes such as joint surgery within the prior two years were excluded. Limited polypharmacy was permitted; healthy older adult subjects were excluded if they took more than four prescription medications, fall-prone subjects were excluded if they took more than six prescription medications. Any subject taking medications known to affect the central nervous system, e.g., sleep aids, anti-seizure, anti-depressant, anti-anxiety medications, etc. was also excluded.

Healthy older adult subjects had no history of falls or near falls and no decline in functional status within the past year. Further, eligibility for the healthy older adult group was limited to older adults who were functionally completely independent in the community without an assistive gait device, engaged in daily physical activity such as stair-climbing, shopping or gardening, and participated at least three times weekly in moderate exercise such as tennis, swimming, walking or running.

Fall-prone subjects had a history of one or more unexplained falls within the past year (range 1 – 6, mean = 3); most also reported numerous near-fall incidents and noticeable decline in their balance requiring adaptations in their daily functional activities. Typical adaptations included the use of a cane when walking outdoors, inability to ascend or descend stairs without a railing, and cessation of balance intensive activities such as dancing or travel. Elderly individuals who were determined to be eligible for the fall-prone group based on questionnaire results subsequently underwent clinical neurologic screening to ensure adequate mental status and vision, sufficient lower extremity somatosensation, and to rule out bilateral vestibular loss (DeMyer, 1994; Reese, 1999; Herdman, 2005).

All subjects received written and verbal descriptions of and instructions for the test procedures. Written consent was obtained from all subjects according to the guidelines proscribed by the Internal Review Board at the University of Maryland before beginning the experiment.

2.2 Experimental set-up

The experimental paradigm, pictured in Figure 2-1, is described more fully in earlier studies (Jeka, Oie et al., 2000; Oie, Kiemel, and Jeka, 2002). Subjects were tested in a ‘multi-sensory moving room’ that permits manipulation of visual and touch (somatosensory) stimuli. They stood on a forceplate (Kistler Instrument Corp., Amherst, NY) in a modified tandem stance position to induce medio-lateral instability. Our goal in inducing instability was to heighten the demand for sensory information for postural control, and tandem stance is an effective way to induce instability on a static, level

surface. Further, deficiencies in postural control in older subjects are greater in the medial-lateral direction (Brauer, Burns and Galley, 2000; Maki, Edmondstone and McIlroy, 2000; Maki, Holiday and Topper, 1994). While on the forceplate the subjects viewed a visual stimulus projected onto a screen. The forceplate recorded center-of-pressure data which are not reported here as our focus was to compare this data set with that of previous COM trajectories of healthy young subjects (Oie, Kiemel, and Jeka, 2002). The visual stimulus was a random pattern of dots (0.2 degrees in diameter) rear-projected on a translucent screen (2.5 m x 1.0 m) located 40 cm in front of each subject. To ensure that the edges of the screen could not be seen and used as stable referents, subjects wore goggles that limited their vision to approximately 100 degrees high and 120 degrees wide. Subjects held their right index fingertip lightly touching the center of a small touch plate, consisting of a smooth, circular metal plate (5 cm in diameter) mounted on a linear positioning table (Daedel 808000 series) supported by a tripod. A computer-controlled servomotor (Compumotor OEM670T) was attached to the positioning table to control motion of the touch plate. The servomotor was programmed with a PID controller to follow a input voltage which approximated a sinusoid using Motion Architect software (Compumotor, Inc). The apparatus was positioned such that motion of the plate was in the medial-lateral plane of the subject at a height approximately level with the hip joint. Three strain gauges (Interface, model MB-5) mounted to the plate allowed transduction of the medial-lateral (F_x), anterior-posterior (F_y) and vertical forces (F_z) applied to the plate. The strain gauges were calibrated in units of force (Newtons) and a computer-controlled auditory alarm sounded when applied forces exceeded 1 N ($\approx 100g$) to insure that the somatosensory input was sensory and not mechanical in nature. Touch plate

displacement was measured using a precision encoder attached to the end of the servomotor. The encoder produced 1000 pulses per revolution. A custom circuit monitored the motor's direction and counted the number of encoder pulses to enable D/A conversion at a resolution of 0.0004 cm. Subjects were continuously monitored during the trials by a 'spotter' who ensured that the subjects' fingertip remained in contact with the center of the moving touchplate such that the subjects' finger/hand/arm moved with the oscillating touchplate. Both the visual and touch stimuli were oscillated in the medial-lateral direction.

Medial-lateral center-of-mass (COM) and head position were recorded at a sampling rate of 50 Hz. using an ultrasound position tracking system (Logitech, Inc.). Two posterior tracking sensors were separately attached to a headband and a neoprene belt at the level of the subject's fourth or fifth lumbar vertebrae to provide approximate positions of the head and COM, respectively. Head trajectory data are not reported here as the changes across conditions in head gains to vision and touch motion stimuli were nearly identical to the changes across conditions observed in COM gains. Moreover, we are interested in the head trajectory data for the sake of developing a multicomponent model of posture, which is the subject of future research. All older subjects wore a snug harness attached to the ceiling loosely with two straps to permit postural sway without sensory cues but prevent a fall to the ground should balance loss occur. Data were collected on a Sun SPARCstation5 workstation using custom LabVIEW data acquisition software and a National Instruments A/D board (SB-MI06).

2.3 Experimental design

Clinical screening for the fall-prone subjects occurred at least 48 hours but not more than one week before the laboratory testing. The battery of clinical screening tests included: the Mini-Mental Status Exam for cognition; lower extremity somatosensory tests including light touch (Semmes-Winstein monofilaments), vibration with a 128 Hz tuning fork, and proprioception (no-vision, verbal-report detection of ten small [2-5 degree] up and down motions at the ankle and great toe); Dynamic Visual Acuity and Halmyagi head thrust maneuver to rule out bilateral vestibular loss, the Berg Balance Scale to assess fall-risk; and the Sensory Organization Test and Limits of Stability Test (NeuroCom International, Inc.) to evaluate the use of sensory inputs for postural control and the volitional control of the center-of-gravity, respectively. The duration of the clinical screening procedures was approximately two hours.

The visual display observed by all subjects was oscillated medial-laterally with approximately sinusoidal motion at 0.20 Hz, while simultaneously the touchplate oscillated medial-laterally with approximately sinusoidal motion at 0.28 Hz. The frequency of motion was held constant for each separate stimulus. The peak amplitudes of the two motion stimuli were varied in five conditions (touch amplitude (mm): visual amplitude (mm)): 8:2, 4:2, 2:2, 2:4, 2:8.

Subjects performed three trials in each of the five conditions for a total of 15 trials. Trials for the healthy and fall-prone older adults were 120 s in duration and the order of trials was pseudo-randomized within five-trial blocks. Seated rests of 120 s were taken between trials, with subjects permitted longer rests if desired. The entire experiment took about three hours to complete.

2.4 Analysis

The time series of the estimated medial-lateral COM motion and the positions of the visual and touch stimuli were recorded (Figure 2-2 A-C). For each of these signals, the mean was subtracted, the signal was zero padded to 2^{20} points, and the Fourier transform was computed. The amplitude spectrum (amplitude as a function of frequency) was computed by taking the absolute value of the Fourier transform (Figure 2-3 3A-E). For each sensory modality the transfer function (frequency-response function) at the stimulus frequency was calculated by dividing the transform of the COM by the transform of the stimulus. (The peak in the stimulus amplitude spectrum was used to accurately determine the stimulus frequency.) Because the visual and touch stimuli were each presented at a different and distinct frequency, the transfer function from each modality in each trial was used to determine the separate contributions of vision and touch to subjects' postural responses.

Gain and phase: Gain and phase at the stimulus frequency were then calculated for both vision and touch modalities. Gain was computed as the absolute value of the transfer function. Phase was computed as the argument of the transfer function and converted to degrees. Gain is a ratio of the amplitude of the response to the amplitude of the stimulus at the driving frequency, and reflects the coupling of postural sway to the stimulus motion. If the response and stimulus amplitudes at the driving frequency are the same, the gain will equal one. The terms visual gain and touch gain refer to COM gain relative to the visual stimulus and COM gain relative to the touch stimulus, respectively. Phase is a measure of the temporal relationship between postural sway and stimulus

motion; postural sway may lead the stimulus (positive values) or lag behind it (negative values).

Position and velocity variability: Position and velocity variability of postural sway are the standard deviation of the residual COM displacement and its derivative, respectively, after removal of the postural sway response at the sensory drive frequencies (cf. Jeka et al., 2000). The residual COM displacement was computed by subtracting sinusoids corresponding to the COM Fourier transform at the touch and vision stimulus frequencies. The derivative of the residual COM trajectory was computed by finite differences using every fifth value of the trajectory, which corresponds to a time step of 0.1 s.

2.5 Statistical analysis

The analysis was to determine within-group, between-condition differences in vision and touch gain, vision and touch phase, position variability and velocity variability, as well as differences between groups. A Group by Condition (3x5) repeated measures MANOVA followed by post-hoc multiple pair wise comparisons with Tukey's correction was used to examine the effect of changing stimulus motion amplitude on gain and phase for both vision and touch, as well as position and velocity variability, in all groups. Subsequently, four separate repeated measures ANOVAs with Bonferroni correction were used to determine whether intra- and inter-sensory re-weighting to vision and touch stimuli had occurred.

3. Results

Typical center of mass (COM) trajectories from exemplar subjects in each group with the concurrent visual and touch motion stimuli trajectories are illustrated in Figure 2-2 A-C. Amplitude spectra for both stimulus trajectories and the three exemplar COM trajectories are shown in Figure 2-3 A-E. The significant amplitude peaks at both 0.20 and 0.28 Hz in the amplitude spectrum of each COM trajectory clearly reflect the simultaneous effects of both the visual and touch motion stimuli. Gain values are derived from the ratios of the COM amplitudes to the motion stimulus amplitudes at each of the two driving frequencies, vision gain at 0.20 Hz and touch gain at 0.28 Hz.

The overall multivariate analysis from the repeated measures MANOVA demonstrated a significant main effect for condition ($p < 0.001$; within subjects), but no significant main effect for group ($p = 0.218$; between subjects) and no group by condition interaction ($p = 0.800$).

3.1 Center of Mass Gain

Figure 2-4 A-B presents the mean COM gains to vision and touch motion stimuli for the three groups, which all showed similar gain values and patterns of response across conditions indicating intra-modality and inter-modality re-weighting between touch and vision. Intra-modality re-weighting is indicated by changes in touch (vision) gain when touch (vision) stimulus amplitude changes. For example, touch gain increases from condition 8:2 to condition 4:2 in Figure 2-4 B. This is due to a decrease in touch stimulus amplitude (from 8 to 4 mm) across these conditions, reflecting *intra-modality* re-

weighting. Across these same conditions, a decrease in vision gain is observed in 2- 4 A, even though visual stimulus amplitude remains constant at 2 mm. This change in vision gain is thus attributable to the change in touch stimulus amplitude, reflecting *inter-modality* re-weighting. Similar results are observed across other conditions.

Changes across conditions: The overall condition main effect emerged from increases in mean visual and touch gain as the amplitude of visual and touch stimulus motion decreased, respectively. Univariate tests showed the changes in gain to be significant across the five conditions for both vision ($p < 0.001$) and touch ($p < 0.001$).

Intra-sensory and inter-sensory re-weighting: To test for intra-sensory and inter-sensory re-weighting for each stimulus modality, four repeated measures analyses were subsequently performed with pair wise comparisons of selected condition results organized by modality. Intra-sensory re-weighting to visual stimuli and inter-sensory re-weighting to touch stimuli were examined using vision and touch gains respectively from the three conditions where the visual stimulus is changing but the touch stimulus is not (touch:vision 2:8, 2:4, 2:2). Intra-sensory re-weighting to touch stimuli and inter-sensory re-weighting to vision stimuli were examined using touch and vision gains respectively from the three conditions where the touch stimulus is changing but the vision stimulus is not (touch:vision 2:2, 4:2, 8:2).

Significant changes in vision gain were found when the visual stimulus amplitude was changing but the touch stimulus amplitude was not (touch:vision 2:2, 2:4, 2:8), and when the touch stimulus amplitude was changing but the vision stimulus amplitude was not (touch:vision 2:2, 4:2, 8:2). The postural sway response to the visual motion stimulus thus reflected both intra- and inter-sensory re-weighting. Intra-sensory re-weighting to

vision was evident between conditions (touch:vision) 2:2 vs. 2:4, 2:2 vs. 2:8, and 2:4 vs. 2:8, (all $p < 0.001$). Inter-sensory re-weighting to vision was evident between conditions 2:2 vs. 8:2 and 4:2 vs. 8:2 (both $p < 0.001$), but not between conditions 2:2 vs. 4:2 ($p = 0.273$).

Significant changes in touch gain were found when the touch stimulus amplitude was changing but the vision stimulus amplitude was not (touch:vision 2:2, 4:2, 8:2) and when the vision stimulus amplitude was changing but the touch stimulus amplitude was not (touch:vision 2:2, 2:4, 2:8). The postural sway response to the touch motion stimulus also reflected both intra- and inter-sensory re-weighting. Intra-sensory re-weighting to touch was evident between conditions (touch:vision) 2:2 vs. 4:2, 2:2 vs. 8:2, and 4:2 vs. 8:2 (all $p < 0.001$). Inter-sensory re-weighting to touch was evident between conditions (touch:vision) 2:2 vs. 2:4 ($p = 0.024$), 2:2 vs. 2:8 ($p = 0.001$) and 2:4 vs. 2:8 ($p = 0.044$).

Group differences: No significant differences in the absolute levels of COM gain relative to both vision and touch motion stimuli were found between groups.

3.2 Phase

Figure 2-5 A-B presents the mean phase relationships of center of mass (COM) to the visual and touch motion stimuli for the three groups. For both vision and touch stimuli, phase was essentially constant (within a -20° to $+30^\circ$ range) across conditions for all groups. An exception is that touch phase in the (touch:vision) 8:2 condition was significantly higher than in the three conditions where the touch stimulus was smallest (2:2, 2:4, 2:8), with all $p \leq 0.046$.

Group differences: While the overall multivariate analysis demonstrated no significant main effects for group, planned post-hoc tests of between subjects effects found marginally significant phase differences between groups for vision but not touch motion stimuli. The COM-Vision phase values were higher in both older adults groups than in the young group ($p \leq 0.041$), but the older groups did not differ from each other ($p = 0.998$). For both older adult groups, the positive phase indicated that the COM tended to lead the visual motion stimulus in all conditions. The young group showed a phase closer to zero degrees. The phase relationship between the COM and the touch stimuli did not differ between groups (all $p \geq 0.912$).

3.3 Sway Variability

Figure 2-6 A-B shows how both position and velocity variability tended to be low when stimulus motion amplitudes were low and increased when either amplitude increased. Position and velocity variability each demonstrated significant changes across condition ($p < 0.001$). Position variability was significantly higher in the (touch:vision) 2:8 condition than in the two conditions with smaller visual stimulus amplitudes (2:4 and 2:2), $p = 0.002$ and $p = 0.019$, respectively. Position variability was also significantly higher in the touch:vision 8:2 condition than in the 2:4 condition ($p = 0.027$).

Velocity variability demonstrated a more pronounced U-shaped curve across conditions than position variability. Velocity variability in the two conditions with a large stimulus amplitude (touch:vision 8:2 and 2:8) did not differ significantly ($p = 0.571$), but each was significantly higher than in every other condition (4:2, 2:2, 2:4), with all $p \leq 0.006$. In the touch:vision 2:2 condition, velocity variability was significantly

lower than in the 4:2 condition ($p = 0.030$), but was not significantly different than in the 2:4 condition ($p = 0.062$).

Group differences: No significant differences in position or velocity variability were found between groups.

4. Discussion

Multi-sensory re-weighting is widely thought to be degraded in older adults relative to young adults, and assumed to be particularly deficient in unstable, fall-prone elders relative to their healthy, age-matched counterparts (Horak, Shupert, and Mirka, 1989). Our results did not support the assumption that the multi-sensory re-weighting adaptation process is deficient in healthy and fall-prone older adults, given sufficiently intact peripheral sensation. Both older adult groups had similar overall levels of vision and touch gain to those found with healthy young adults (Oie, Kiemel, and Jeka, 2002). In all groups, vision gains increased as the visual motion stimulus amplitude decreased, and touch gains increased as the touch motion stimulus amplitude decreased, reflecting intact intra-modality re-weighting.

Healthy elderly and fall-prone groups also showed intact inter-sensory re-weighting to vision and touch. Inter-sensory re-weighting of vision was evidenced by changes in vision gain when the touch stimulus amplitudes changed while the visual stimulus amplitudes remained constant. Inter-sensory re-weighting of touch was evidenced by changes in touch gain when the vision stimulus amplitudes changed while the touch stimulus amplitudes remained constant. These results provide evidence that

postural instability in older adults and the fall-prone elderly with sufficiently intact peripheral sensation is not due to the inability to adaptively reweight visual and touch information.

Our results suggest a relationship between postural sway at the stimulus frequencies, which is described by the visual and touch gains (Fig. 4), and postural sway at other frequencies, which is described by sway variability (Fig. 6). Variability tended to increase when either the visual or touch stimulus amplitude increased. An adaptive postural model from Carver, Kiemel et al. (2005) reproduces this increase in variability with stimulus amplitude (Jeka, Allison et al., 2006). In the model, sensory weights are adjusted to minimize the mean-squared ankle-muscle torque specified by the neural controller. When the body leans relative to vertical, more muscle torque is needed to counteract the force of gravity. As a result, in quiet stance (stationary visual scene and touch plate) minimizing mean-squared muscle torque is roughly the same as minimizing variability. When the visual scene and touch plate move, power appears at the stimulus frequencies and the system re-weights sensory inputs to reduce this power. However, there is a trade-off. Reducing power at the oscillation frequencies leads to increased power at other frequencies, so re-weighting leads to an increase in sway variability.

Age Effects in the Existing Literature

These results differ from previous studies which demonstrated larger postural sway responses to sensory stimulation in the healthy and fall-prone elderly (Sundermier, Woollacott, Jensen, and Moore, 1996; Perrin, Jeandel et al., 1997; Borger, Whitney et al., 1999; Loughlin and Redfern, 2001). Though none of these studies focused on the question

of multi-sensory re-weighting, all showed an age effect, that is, the same visual stimuli produced greater postural sway in healthy older adults compared to healthy young adults. Sundermier and colleagues (Sundermier, Woollacott, Jensen, and Moore, 1996) further found that the fall-prone older adults had greater postural sway than both healthy older and healthy young adults.

While not explicitly mentioned, Borger and colleagues (Borger, Whitney, Redfern, and Furman, 1999) indirectly explored inter-sensory re-weighting to visual stimulus motion in healthy young versus older adults. They varied the amplitudes of the visual motion stimulus at two frequencies on both a fixed surface and a sway-referenced surface. Though these authors did not report vision gains at the visual driving frequencies, we obtained roughly comparable estimates of the vision gains by (1) converting the visual motion stimulus amplitudes at 0.25 Hz from degrees to centimeters (assuming an eye-level height of 170 cm) and (2) dividing the postural sway response (peak to peak amplitude of the COP) by the visual motion stimulus amplitude. For both age groups, at the same frequency and amplitudes of visual stimulus motion, vision gains were higher in the sway-referenced surface condition, when somatosensory cues were inaccurate, than in the fixed surface condition when somatosensory cues were accurate. This increase in visual gain is interpretable as inter-sensory re-weighting of visual information in healthy young and older adults, a finding consistent with our results. Sway-referencing represents a severe alteration in sensory information that clearly leads to re-weighting of available sensory modalities to compensate for the loss of accurate sensory information. The current method of independently changing the relative amplitude of different modalities illustrates that the central process of re-weighting is

sensitive to very small changes in the sensory environment. The precise nature of the re-weighting process means that the nervous system is not just re-weighting in response to severe changes in sensory input, but may be continually re-weighting as we move through changing environments.

One explanation for the discrepancy between our results and previous studies is that substantially different methods and measures were used. First, the duration of trials was much longer in this study than was typically seen in earlier studies. We used 120 second trials versus a range of 10 to 90 second trials in other studies. Several of the cited studies used the standardized Sensory Organization Test[®] (SOT) which has three 20 second trials in each sensory condition. Although little is known about the time scale of multi-sensory re-weighting, previous studies estimate the time-course of re-weighting in young adults to be on the order of 5 – 30 sec (Dijkstra, Schonert et al., 1994; Peterka and Loughlin, 2004). Teasdale and colleagues have reported that adaptation to changes in sensory conditions in older adults is slower than in young adults (Teasdale, Stelmach, Breunig, and Meeuwsen, 1991b; Teasdale, Bard, LaRue, and Fleury, 1993; Hay, Bard et al., 1996; Teasdale and Simoneau, 2001). It may be that longer exposure to the sensory conditions in this study permitted successful multi-sensory re-weighting in our elderly subjects that might not have occurred with much shorter trials. We examine this hypothesis directly in a subsequent study (Allison, Kiemel & Jeka, 2005).

Second, compared to other studies in which dynamic sensory stimuli were large and/or sudden (c.f. Sundermier et al., 1996; Simoneau et al., 1999), the dynamic sensory stimuli provided in this study were small in amplitude and oscillated slowly. A follow-up study using both small and large amplitude visual stimuli may help to determine whether

differences in stimulus parameters might account for our different result (Allison, Kiemel & Jeka, 2005).

Lastly, the dynamic somatosensory stimulus in the current study was provided using the fingertip touch plate and not a moving support surface (e.g. in the SOT) which introduces biomechanical changes as well as sensory manipulation. The ‘gentle’ conditions in this study did not physically destabilize the subjects and thus minimized non-sensory biomechanical, force and timing demands that if present might have resulted in age-related differences in postural sway.

Visual Dependence in Older Adults

Several studies that have demonstrated increased postural sway in response to dynamic visual environments in healthy and fall-prone older adults have concluded that these older adults are “visually dependent” or “visually sensitive” (Wade, Lindquist et al., 1995; Sundermier, Woollacott, Jensen, and Moore, 1996; Simoneau, Teasdale, Bourdin, Bard, Fleury, and Nougier, 1999). Visual dependence is conceived of as an over-reliance on visual cues that may be inaccurate or unreliable in the presence of stable and reliable somatosensory and vestibular cues. It is considered a multi-sensory re-weighting deficit in that there is a failure to ‘switch’ from inaccurate visual information to accurate somatosensory and vestibular information, i.e., to down-weight vision and up-weight somatosensation and vestibular inputs.

The favored explanation for visual dependence is age- or disease-related peripheral sensory loss in the somatosensory and vestibular systems. If the postural control system cannot rely on somatosensory and/or vestibular inputs, it must over-rely

on vision. Results from Peterka (Peterka, 2002) support this concept. He found that subjects with known bilateral vestibular loss weight vision and proprioceptive cues more highly than do individuals with intact vestibular function.

The healthy and fall-prone older subjects in our study did not demonstrate an over-reliance on vision. Specifically, they did not show larger responses (higher gains) to dynamic visual stimuli compared to healthy young adults. Nor did they produce smaller responses (lower gains) to dynamic touch stimuli that might have reflected an under-reliance on somatosensation. Together these results do not reflect a compensatory ‘upweighting’ of visual motion cues in response to a reduced or absent ability to use touch cues. This is perhaps because subjects with known or clinically detectable peripheral sensory loss were excluded from our study. If peripheral somatosensory and vestibular inputs are adequately detected, no compensatory increase in the relative reliance on visual inputs is necessary.

We did observe age-related differences in the vision phase, however. For both older adult groups, the phase was higher than that of the young adult group in four of the five conditions. For the older adults, postural sway led the visual stimulus by 20 to 30 degrees, while the young adults demonstrated either zero phase shift or a slight phase lag. A phase lead could be caused by an increase in the stiffness of the postural control system. However, such differences would also lead to a phase lead for the touch stimuli, which was not observed. It is possible that this age-related difference in vision phase reflects some form of visual dependence, but in the absence of higher vision gains we consider this explanation unlikely. At present we have no satisfactory explanation for this result.

Several recent studies have found larger responses to dynamic visual stimuli despite the employment of careful screening tests for peripheral somatosensory and vestibular sensory loss and the exclusion of subjects with sensory loss (Borger, Whitney, Redfern, and Furman, 1999; Simoneau, Teasdale, Bourdin, Bard, Fleury, and Nougier, 1999). Their findings argue against peripheral somatosensory or vestibular loss as the sole or primary causative factors of “visual sensitivity” in healthy and fall-prone older adults. Differences between studies in the sensory stimuli parameters, measurement and analysis methods, and trial duration may contribute to the discrepant findings.

Conclusions

In the current study we provide new evidence that healthy and fall-prone older adults with sufficiently intact peripheral sensation are capable of sensitive intra- and inter-sensory re-weighting to small amplitude visual and touch motion stimuli in much the same manner as healthy young adults. Further investigation of multi-sensory re-weighting in the elderly using mechanism-specific measures and methods across a range of stimulus parameters is warranted to resolve discrepancies between results from the current study versus prior studies. The time scale of multi-sensory re-weighting and how it may change with age and disease is unknown. If older adults take longer to successfully reweight multi-sensory stimuli than young adults, results from studies with shorter trial durations may reflect slowed processing rates versus actual re-weighting deficits.

Figure Captions

FIGURE 2-1

Experimental set-up.

A young subject is illustrated in a modified tandem stance facing the computer generated visual display while lightly contacting the touch surface with her right fingertip. Markers on the occiput and lower lumbar region tracked the trajectories of the head and estimated center of mass, respectively.

FIGURE 2-2 A – C

Time series.

Three exemplar time series are shown from (A) a fall-prone older adult, (B) a healthy older adult, and (C) a healthy young adult. A 60-s segment of data shows the time series of visual display motion at 0.2 Hz (upper trace), medio-lateral center of mass displacement (middle trace), and touch surface motion at 0.28 Hz (lower trace). In these exemplars, visual motion amplitude is 2mm and touch surface motion amplitude is 4mm.

FIGURE 2-3 A – E

Amplitude spectra.

Amplitude spectra for (A) visual display motion, (B) touch surface motion, and center of mass displacements from the same (C) fall-prone older adult, (D) healthy older adult, and (E) healthy young adult shown in the exemplar trials of Fig. 2.

FIGURE 2-4 A – B

Group mean gain results.

The mean (\pm S.E.) (A) visual gain and (B) touch gain versus condition for each group. For vision gains, intra-sensory re-weighting is reflected by changes in vision gain between conditions 2:2, 2:4, and 2:8. Inter-sensory reweighting is reflected by changes in vision gain between conditions 2:2, 4:2, and 8:2. For touch gains, intra-sensory re-weighting is reflected by changes in touch gain between conditions 2:2, 4:2, and 8:2. Inter-sensory reweighting is reflected by changes in touch gain between conditions 2:2, 2:4, and 2:8.

FIGURE 2-5 A – B

Group mean phase results.

The mean (\pm S.E.) (A) visual phase and (B) touch phase versus condition for each group.

FIGURE 2-6 A – B

Group mean position and velocity variability results.

The mean (\pm S.E.) (A) position variability and (B) velocity variability versus condition for each group.

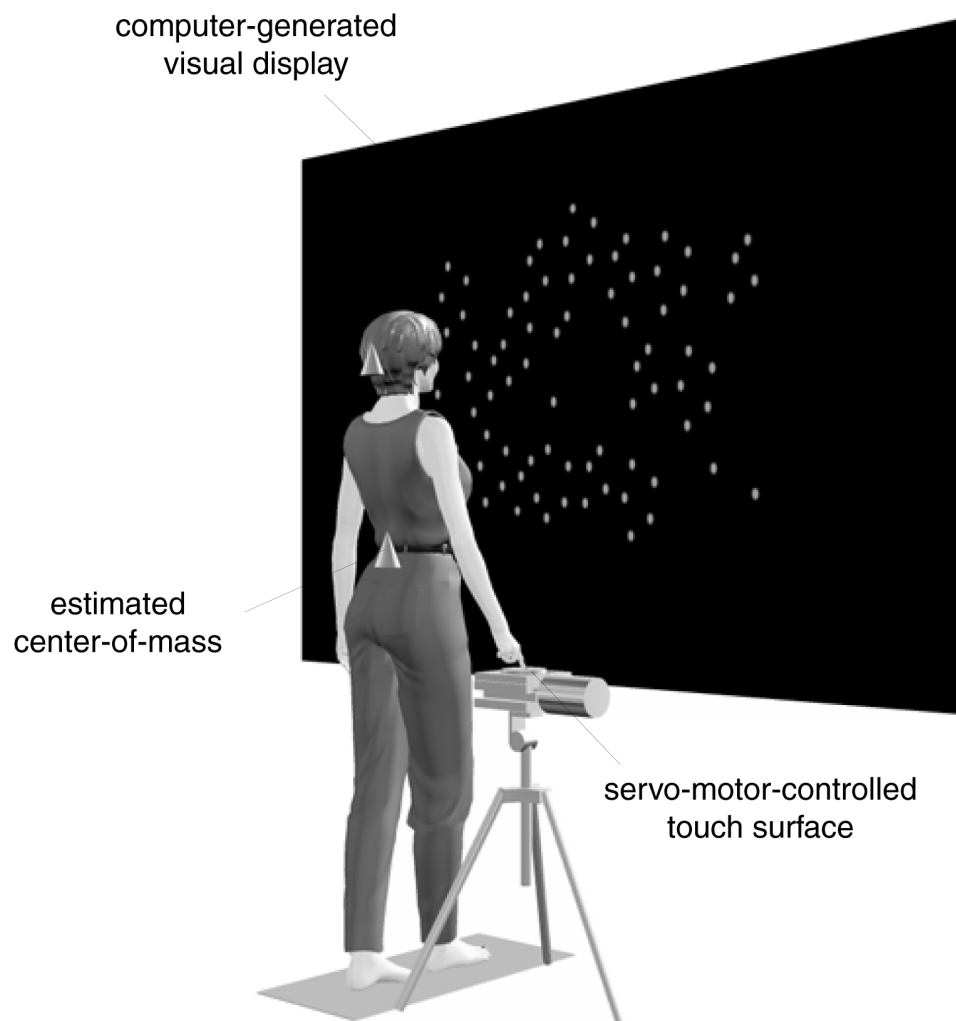


Figure 2-1 Two-frequency vision and touch experimental paradigm.

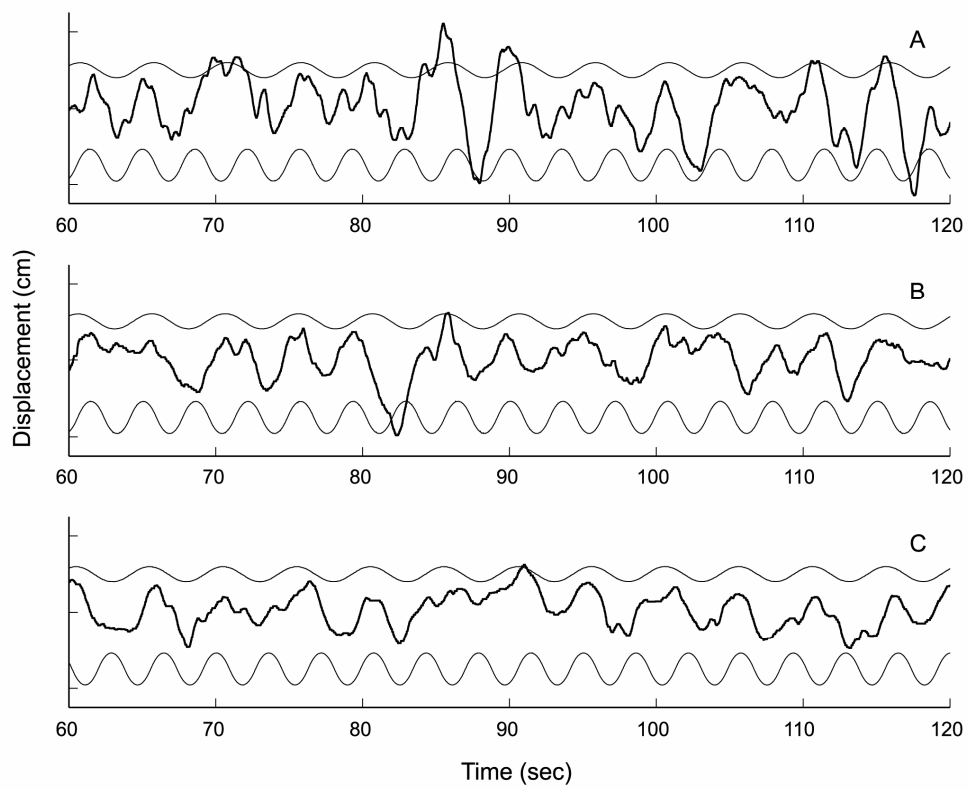


Figure 2-2: Exemplar time series

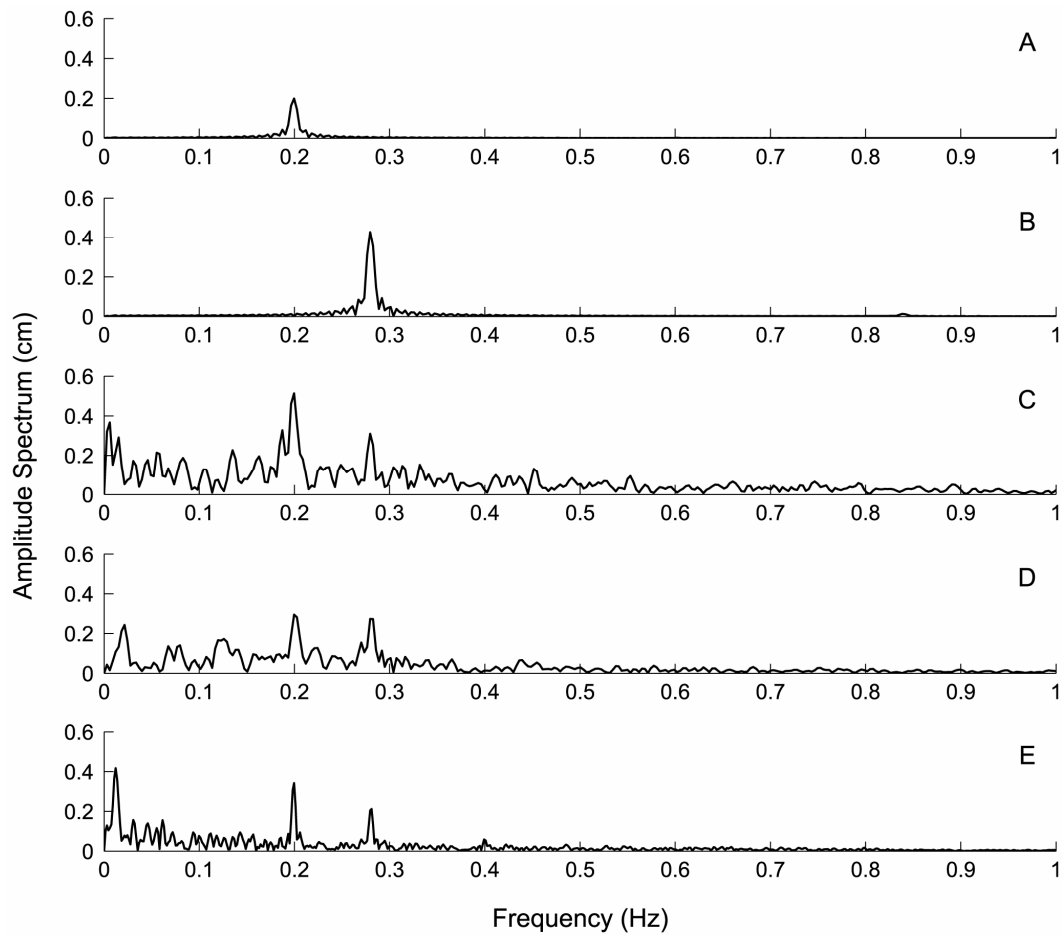


Figure 2-3: Exemplar amplitude spectra

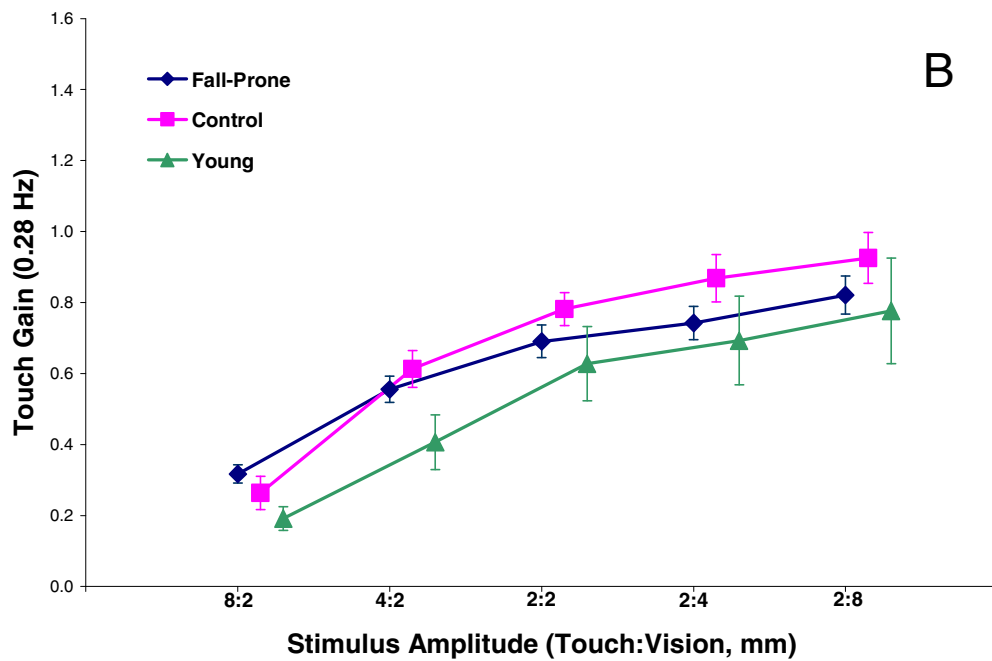
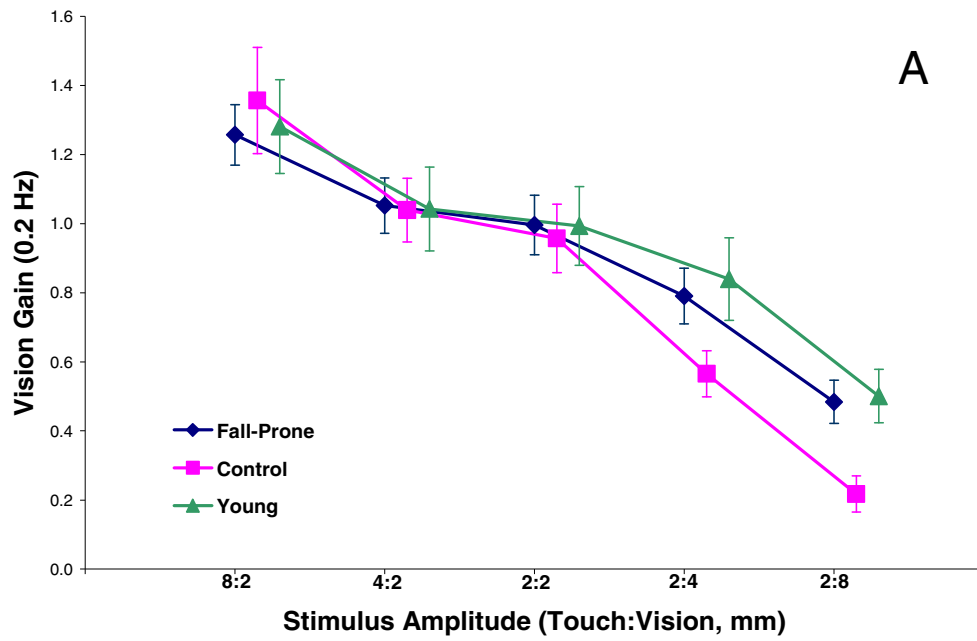


Figure 2-4: Vision and Touch Gains

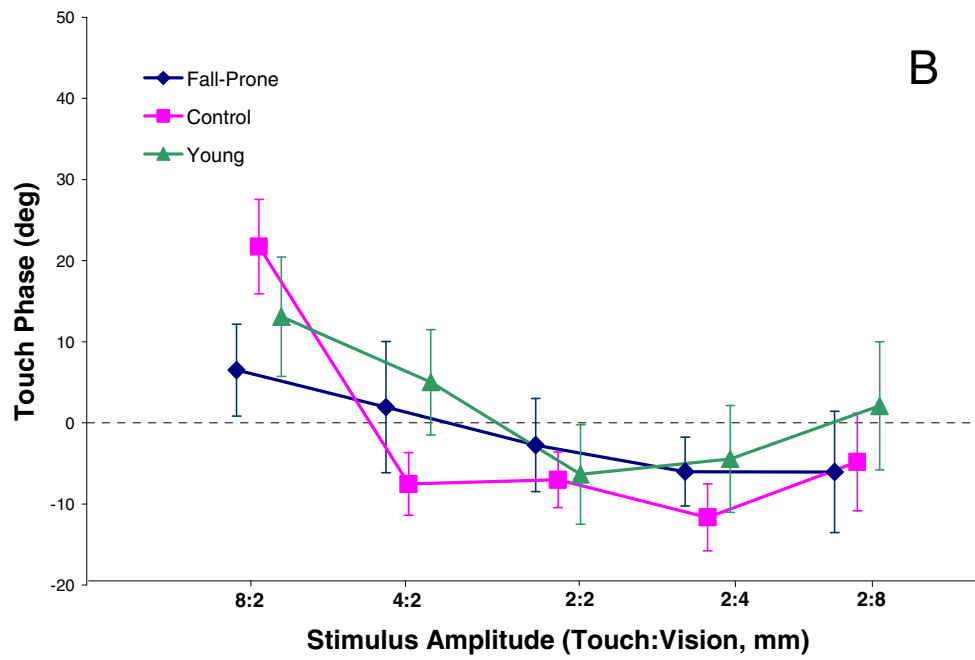
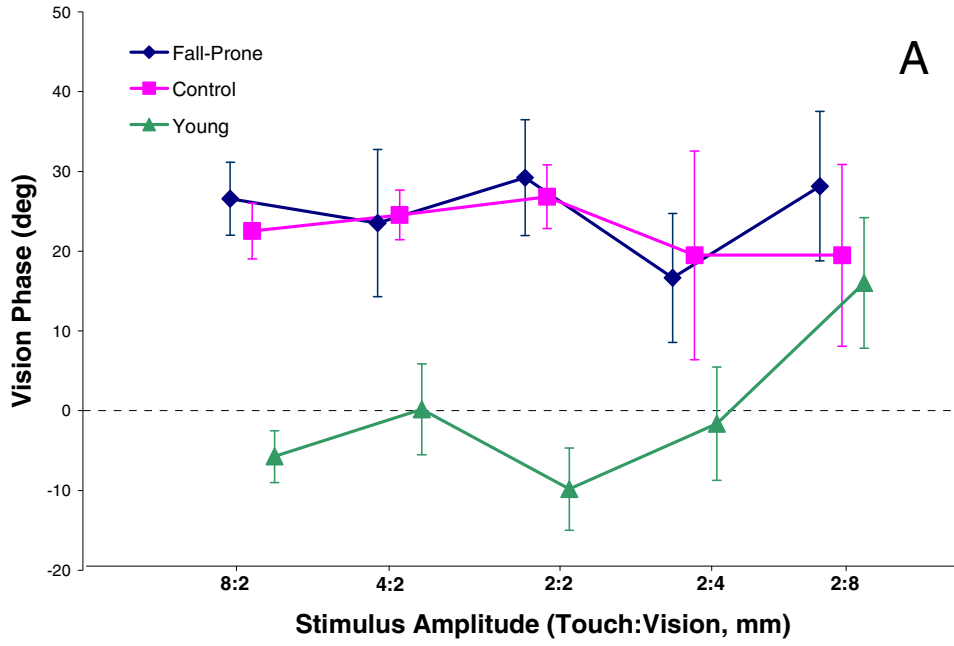


Figure 2-5: Vision and Touch Phases

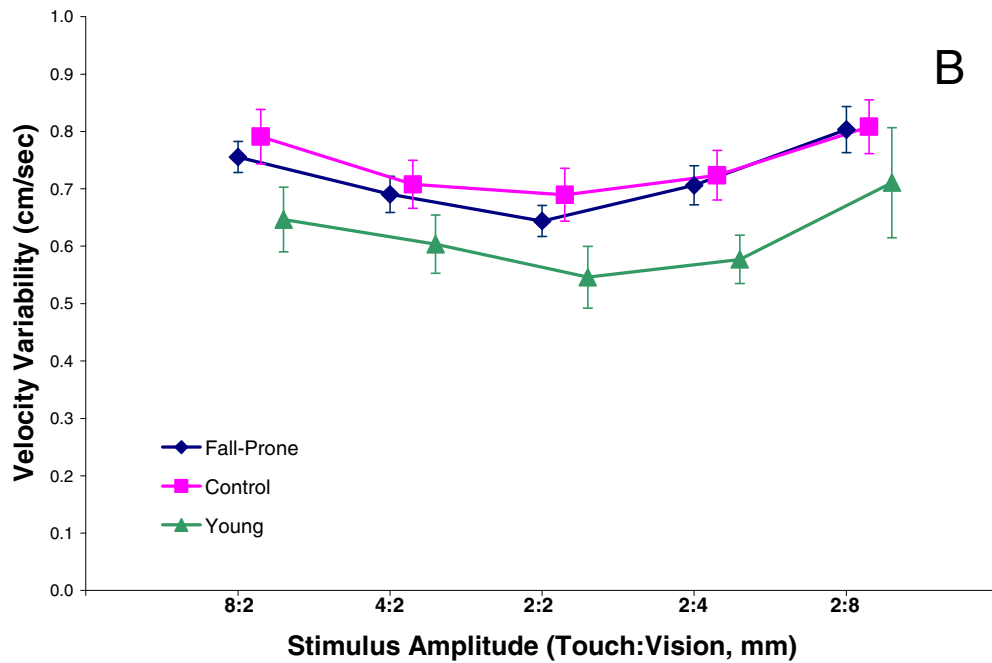
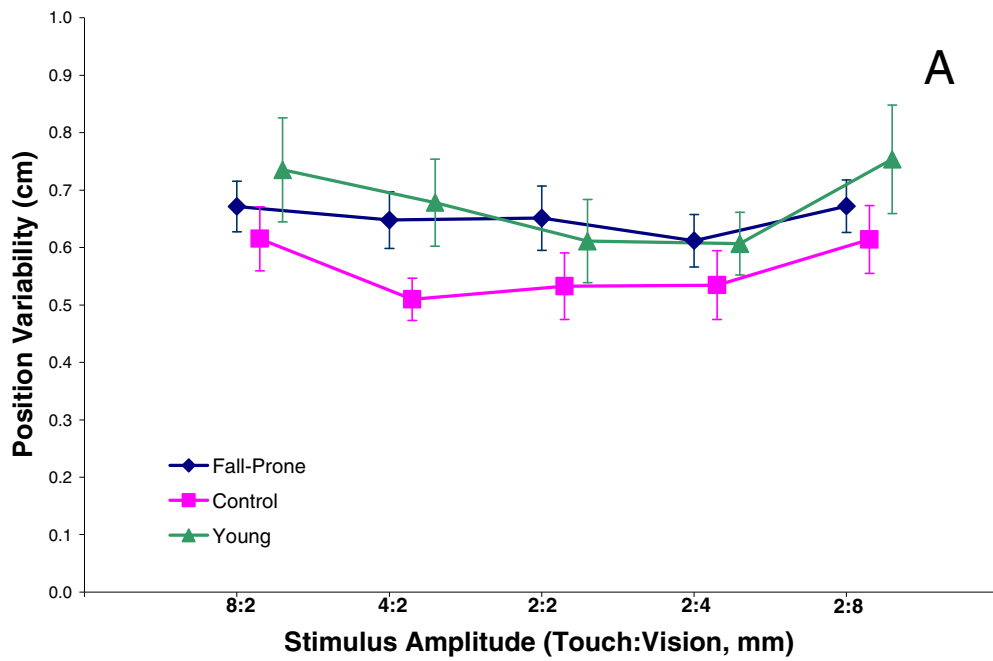


Figure 2-6: Position and Velocity Variability

Chapter 3:

The Effect of Sensory-challenge Balance Exercises on Multi-sensory Re-weighting and Balance in Fall-Prone Older Adults

1. Introduction

When environmental conditions change, the postural control system includes an adaptive process that prioritizes the visual, vestibular and somatosensory inputs that are providing the functionally most reliable information (i.e., multi-sensory re-weighting). Though poorly understood, multi-sensory re-weighting is generally held to be impaired in older adults and more so in the fall-prone versus healthy elderly (Horak, Shupert et al., 1989; Woollacott, 2000). A compromised ability to update and prioritize sensory information would impair the determination of body position and motion in space. This, in turn, might affect perceptual recognition of the need for balance responses, and the accuracy of neural commands to the musculature for the production of effective balance responses. Deficient multi-sensory re-weighting may thus contribute to imbalance and falls in older adults.

In the current study we targeted multi-sensory re-weighting as a specific mechanism underlying flexible balance control. Prior balance exercise intervention studies that included exercises targeting one or more of the sensory systems (vision, vestibular, and somatosensory) have typically reported improvements in balance and a reduction in fall-risk (Hu and Woollacott, 1994; Shumway-Cook et al, 1997; Rose and Clark, 2000). In a meta-analysis of balance exercise interventions, Whipple found that one characteristic of successful exercise interventions was the incorporation of sensory challenge activities (Whipple, 1997). We hypothesized that following a sensory-

challenge exercise program, we would see improvements in clinical measures of balance and fall risk. Our intent was to: 1) target a specific mechanism (i.e., multi-sensory re-weighting) underlying flexible balance control and; 2) precisely measure whether sensory challenge exercises could influence this mechanism.

Here we use an analysis that provides separate quantitative measures of postural responses to visual and somatosensory cues. Using this method we have previously demonstrated changes in gain interpreted as multi-sensory re-weighting in healthy young adults, healthy older adults, and fall-prone older adults (Allison, Kiemel & Jeka, 2006; in press).

2. Methods

2.1 Subjects

Older adults (70 years and older) with a self-reported history of falls in the prior year were recruited from a congregate retirement community. Volunteers completed an eligibility questionnaire and were excluded if they had any of the following: visual conditions affecting daily function (e.g. reading, driving, etc.); dizziness, vertigo or vestibular conditions; lower extremity numbness or conditions such as diabetes leading to somatosensory loss; neurological disorders; cognitive decline; recent musculoskeletal changes, or insufficient endurance. Subjects were excluded if they took more than six prescription medications or any medication known to affect central nervous system control of posture, e.g., sleep aids, anti-depressants, etc.

Twenty-eight fall-prone older adults with a mean age of 84 ± 5 years were accepted into the study. Subjects had a history of one or more unexplained falls within the past year (range 1 – 6, mean = 3). All subjects subsequently underwent clinical screening by a licensed physical therapist to ensure adequate mental status, intact lower extremity somatosensation, and to rule out bilateral vestibular loss.

Written consent was obtained from all subjects according to the guidelines proscribed by the Internal Review Board at the University of Maryland.

2.2 Experimental set-up

The experimental paradigm (Fig. 3-1) is described more fully in earlier studies (Jeka, Oie et al., 2000; Oie, Kiemel et al., 2002). Subjects were tested in a ‘multi-sensory moving room’ that permits manipulation of visual and touch (somatosensory) stimuli. They stood on a forceplate (Kistler Instrument Corp., Amherst, NY) in a modified tandem stance to induce medio-lateral instability while viewing a visual stimulus. The random pattern visual stimulus was rear-projected on a screen located in front of each subject. To ensure that the edges of the screen could not be seen, subjects wore goggles that limited their vision to approximately 100 degrees high and 120 degrees wide. Subjects held their right index fingertip lightly touching the center of a small forceplate, located at hip height on the right side. The touchplate was instrumented to sound an alarm if the touch force exceeded 1 N to prevent the surface from being used as a mechanical stabilizing force. Both the visual and touch stimuli could be oscillated in the medial-lateral direction.

Medial-lateral center-of-mass (COM) and head position were approximated using an ultrasound position tracking system (Logitech, Inc.). Two posterior tracking sensors were separately attached to a headband and a belt at the level of the subject's fifth lumbar vertebrae. Subjects wore a snug harness attached to the ceiling loosely to permit postural sway without tactile feedback from the harness, but to prevent a fall should balance loss occur. An assistant stood close to each subject. Data were sampled at 50 Hz. Head trajectory data will be reported in a subsequent manuscript.

2.3 Experimental design

Subjects served as their own controls. Clinical and laboratory tests were performed at three different times, pre-test 1, pre-test 2 and post-test. An eight-week “no-training” control period occurred between pretests 1 and 2. An eight-week “training” period occurred between pre-test 2 and post-test.

2.3.1 Clinical testing

Clinical testing was performed by a licensed physical therapist prior to laboratory testing. Clinical screening tests used for inclusion/exclusion purposes included the Mini-Mental Status Exam; lower extremity somatosensory tests including light touch, vibration, and proprioception; as well as the Dynamic Visual Acuity test and Halmyagi head thrust maneuver to rule out bilateral vestibular loss (DeMyer, 1994;Herdman, 2005;Reese, 1999).

Clinical outcome measures included the Activities-specific Balance Confidence Scale [ABC Scale] (Myers, Powell et al., 1996;Powell and Myers, 1995), bilateral lower

extremity range-of-motion (goniometry) and strength (handheld dynamometer; Chatillon CSD100, AMETEK, Inc.); the Berg Balance Scale (Berg, Wood-Dauphinee et al., 1995; Berg, Wood-Dauphinee et al., 1992), the Sensory Organization Test (Jacobson, 1997) and Limits of Stability Test (Clark and Rose, 2001) (NeuroCom International, Inc.). Range of motion and strength were assessed bilaterally for the following joint motions: hip flexion, extension, abduction, adduction; knee flexion and extension; and ankle dorsiflexion and plantarflexion. A summary range-of-motion limitation score was obtained by adding the limitations found at each joint. A summary strength score was obtained by adding the number of pounds of force produced at each of the 16 measured joint motions.

2.3.2 Laboratory multi-sensory re-weighting test

The visual display observed by all subjects was oscillated medial-laterally at 0.20 Hz, while simultaneously the touchplate oscillated medial-laterally at 0.28 Hz. Provision of the visual and touch stimuli at separate frequencies allowed subsequent analysis to distinguish the postural response to each stimulus. The peak amplitudes of the two motion stimuli were varied in five conditions (touch:visual amplitudes (mm)): 8:2, 4:2, 2:2, 2:4, 2:8. Prior research has shown that systematic amplitude variation leads to corresponding changes in postural sway gain (response amplitude divided by stimulus amplitude), which can be interpreted as re-weighting of the sensory modalities (Oie, Kiemel, and Jeka, 2002; Ravaioli, Oie et al., 2005; Peterka & Benolken, 1995). Subjects performed three trials in each of the five conditions for a total of 15 trials. Trials were 120 s in duration and the order of trials was pseudo-randomized within five-trial blocks.

Seated rests of 120 s were taken between trials, with subjects permitted longer rests if desired.

2.3.3 Sensory challenge balance exercise program

Following pre-test 2, subjects attended two, 45-minute exercise sessions each week for eight weeks. Exercise sessions were “one-on-one” with a licensed physical therapist (PT) or physical therapist assistant (PTA) who had been trained in the research exercise protocol.

All exercises were performed on a SMART Balance Master (NeuroCom International, Inc.), a computerized balance testing and training device that permits operator controlled surface and/or visual environment motion and, if desired, provides visual feedback about center-of-gravity position and motion. The purposes of the balance exercises were to improve (1) estimation of body position and motion in space and (2) adaptation to changing sensory environments. Subjects were asked to stand as steadily as they could without stiffening. All subjects followed the same standardized exercise progression, however, the initial difficulty level of the exercises was adjusted for each subject based on their balance abilities. Exercises were made progressively more difficult by modification of (1) stance position, (2) head position/motion, (3) the availability of vision, (4) visual surround motion, (5) the availability of somatosensory information (6) support surface motion and (7) target size.

Initially, continuous and immediate COG visual feedback was provided. Over the first four weeks of balance training, visual feedback was increasingly delayed and provided only after the practice trial was completed and after subjects had been asked to

verbally evaluate their own performance. In this manner subjects were “weaned” away from the extrinsic visual feedback. Summary visual feedback was rarely provided during the second half of the exercise program.

2.4 Multi-sensory re-weighting analysis

Figure 3-2 A-C shows exemplar time series of the medial-lateral COM and the visual and touch stimuli. The amplitude spectrum was computed from the time series by taking the absolute value of the Fourier transform (Fig. 3-3 A-E). For each sensory modality the transfer function at the stimulus frequency was calculated by dividing the transform of the COM by the transform of the stimulus. The transfer function of each modality was used to determine the separate contributions of vision and touch to subjects’ postural responses.

Gain and phase: Gain is a ratio of the amplitude of the response to the amplitude of the stimulus at the driving frequency, and reflects the coupling of postural sway to the stimulus motion. Gain was computed as the absolute value of the transfer function. If the response and stimulus amplitudes at the driving frequency are the same, the gain will equal one. The terms visual gain and touch gain refer to COM gain relative to the visual stimulus and touch stimulus, respectively. Phase is a measure of the temporal relationship between postural sway and stimulus motion; postural sway may lead the stimulus (positive values) or lag behind it (negative values). Phase was computed as the argument of the transfer function and converted to degrees.

Position and velocity variability: Position and velocity variability of postural sway are the standard deviation of the residual COM displacement and its derivative,

respectively, after removal of the postural sway response at the stimulus frequencies (cf. Jeka et al., 2000). The residual COM displacement was computed by subtracting sinusoids corresponding to the COM Fourier transform at the touch and vision stimulus frequencies. The derivative of the residual COM trajectory was computed by finite differences using every fifth value of the trajectory, which corresponds to a time step of 0.1 s.

2.5 Statistical analysis

All statistical analysis were performed using SPSS[®] Version 12. Laboratory re-weighting data and clinical test data were analyzed separately.

2.5.1 Multi-sensory re-weighting

The analysis was to determine between-condition differences in vision and touch gain, and vision and touch phase, at each test period, as well as differences between test periods. A Test by Condition (3x5) repeated measures MANOVA, followed by planned multiple pair-wise comparisons with Bonferroni correction, examined the effect of changing stimulus motion amplitude on gain and phase for both vision and touch at each test period.

Position variability and velocity variability was examined with a Test by Condition (3x5) repeated measures MANOVA. A lack of significant change in position or velocity variability across conditions would indicate that responses were primarily at the stimulus frequencies rather than at all frequencies of sway.

2.5.2 Clinical tests

The ABC Scale and Berg Balance Scale are ordinal rating scales and were analyzed using Friedman's test, with planned multiple comparisons performed using related samples paired T-tests with Bonferroni correction of the significance level to $p \leq 0.016$. Bilateral lower extremity range of motion and strength, the SOT and the LOS tests were examined using a 3 x 3 (Test by Trainer) Repeated Measures MANOVA. Included in the multivariate analysis were: 1) a weighted average of the mean scores (Composite score) in each of the six sensory conditions from the SOT; and 2) the sum of the eight Maximum Excursion scores from the LOS test.

3. Results

Six subjects withdrew from the study prior to the start of training. Twenty subjects finished all 16 sessions. The exercise program was well tolerated with no adverse effects.

3.1 Laboratory multi-sensory re-weighting

Evidence of intact multi-sensory re-weighting was found, reflected by systematic patterns of change in gain values as the motion stimulus amplitudes changed (Fig. 3-4 A & B). This adaptive response was present before and after training, implying that multi-sensory re-weighting in fall-prone elders is not deficient. However, both vision and touch gain values appeared to decrease post-training, indicating a change in the ability to discriminate and dissociate self- versus environmental motion.

A 3 x 5 (Test by Condition) repeated measures MANOVA for vision and touch gain revealed significant main effects for Test ($p = 0.041$) and Condition ($p < 0.001$). A second 3 x 5 (Test by Condition) MANOVA for position and velocity variability revealed only a significant main effect for Condition ($p < 0.001$).

3.1.1 Differences across conditions: Evidence for multi-sensory re-weighting

Vision gain: Intra-sensory and inter-sensory re-weighting of vision were found, supported by significant differences in vision gains for Condition ($p < 0.001$), Fig. 3-4-A. Planned comparisons revealed significant differences between all pairs of conditions ($p < 0.001$), except between Conditions 4:2 and 2:2. Vision gains changed across the three conditions where the visual amplitude was changing (touch:vision 2:2, 2:4, 2:8; *intra-sensory*) and also across the conditions where the visual amplitude was constant while the touch amplitude was changing (touch:vision 8:2, 4:2, 2:2; *inter-sensory*).

Touch gain: Only intra-sensory re-weighting of touch was found, supported by significant differences in touch gains for Condition ($p < 0.001$), Fig. 3-4-B. Planned comparisons revealed highly significant differences between Condition 8:2 and all other conditions (all $p < 0.001$), and Condition 4:2 and all other conditions (all $p \leq 0.001$). Touch gains changed across the three conditions where the touch amplitude was changing (touch:vision 8:2, 4:2, 2:2; *intra-sensory*) but not across the conditions where the touch amplitude was constant while the vision amplitude was changing (touch:vision 2:2, 2:4, 2:8; *inter-sensory*).

Vision and touch phase: Vision phase showed a small (10 to 35 degree) phase lead and touch phase showed a small phase lag across conditions (Fig. 3-5 A & B). Univariate tests for Condition found significant differences in touch phase ($p < 0.001$), but no significant differences in vision phase. Planned comparisons revealed significant differences in touch phase between Condition 8:2 and all other conditions (all $p \leq 0.037$).

Position and velocity variability: Univariate tests found significant differences between conditions for position and velocity variability ($p = 0.002$ and $p < 0.001$, respectively) (Fig. 3-6 A & B). A trend in position and velocity variability toward lower variability when the two sensory motion amplitudes are lowest (2:2) versus higher variability when either amplitude is largest (8:2 and 2:8) is seen. Planned comparisons found significant differences in position variability between Condition 2:2 vs. Conditions 8:2 and 2:8, both $ps \leq 0.009$, and between Condition 2:4 vs. Condition 2:8, $p = 0.02$. For velocity variability, comparisons revealed significant differences between Condition 2:2 and all other conditions (all $p \leq 0.004$). Significant differences were also found between Condition 8:2 vs. Conditions 4:2 and 2:4; Condition 4:2 vs. Condition 2:8, and Condition 2:4 vs. Condition 2:8, all $ps < 0.004$.

3.1.2 Differences between test periods: Evidence for the effect of sensory-challenge balance exercises

Gain and Phase for Vision and Touch: Univariate tests revealed significant decreases between test periods for touch gain and phase ($p < 0.006$ and $p < 0.05$, respectively), but not for vision gain or phase. Though vision gain changed in the same direction as touch gain, the power to detect differences in vision gain and phase was much lower than in touch gain and phase (0.389 and 0.197 vs. 0.861 and 0.604,

respectively). The standard error for vision gain values was over twice that for touch gain values (vision SE = 0.16; touch SE = 0.07), which may explain why the decrease in touch gain values reached significance but the vision gain values did not. Planned comparisons found a significant decrease in touch gain only between pretest 1 and posttest ($p = 0.003$). No significant differences were found between any pairs of test periods for vision gain or phase, or touch phase.

Position and velocity variability: No significant differences in position or velocity variability were found between any of the test periods.

3.2 Clinical tests

3.2.1 ABC Scale and Berg Balance Scale

Berg Balance Scale scores showed an increase across tests: means/SDs for Pretest 1, Pretest 2 and Posttest were 48.7/5.6, 50.4/5.0, and 51.75/3.9. These differences were significant (Freidman test, $p < 0.001$). Planned related-samples paired T-tests with a Bonferroni corrected significance level of $p \leq 0.016$ demonstrated significant differences between pretest 1 vs. pretest 2 ($p = 0.006$), as well as pretests 1 and 2 vs. the posttest ($ps < 0.003$). No significant between-test differences were found for the ABC Scale scores (Freidman test, $p = 0.115$).

3.2.2 Sensory Organization Test, Limits of Stability test, Strength and Range of Motion

Values for the SOT, LOS and Strength showed significant changes as a function of Test, but no differences were observed for the lower extremity range of motion scores ($p = 0.227$). A Test by Trainer (3 x 3) MANOVA found a significant main effect for Test (Wilk's Lambda, $p < 0.001$), but no significant main effect for Trainer, and no significant Test x Trainer interaction. Univariate tests revealed significant between-test differences for the SOT Composite scores ($p < 0.001$; means/SDs for Pretest 1, Pretest 2 and Posttest were 62.7/12.7, 69.2/10.5, and 78.0/8.5), the LOS summary Maximum Excursion scores ($p < 0.001$; means/SDs for Pretest 1, Pretest 2 and Posttest were 59.8/12.8, 61.5/13.3, and 69.3/12.9) and the summary lower extremity strength scores ($p = 0.013$; means/SDs for Pretest 1, Pretest 2 and Posttest were 162.8/41.2, 165.7/41.3, and 180.9/41.4).

Significant between-test differences were found for the SOT Composite score for all three pairs of tests ($ps \leq .005$). For the LOS summary Maximum Excursion score and the summary lower extremity strength score, significant differences were found between pretests 1 and 2 vs. the posttest ($ps \leq 0.05$), but not between pretests 1 and 2.

3.2.3 Individual Sensory Organization Test Conditions

Individual SOT conditions showed changes as a function of Test, supported by a significant between-test difference (Wilk's Lambda, $p < 0.001$). Univariate tests found significant between-test differences for SOT conditions four ($p = 0.023$), five ($p < 0.001$) and six ($p < 0.001$). Pair-wise comparisons for sensory condition four showed a nearly significant difference only between pretest 1 vs. posttest ($p = 0.061$, other pairs $ps > 0.234$). For SOT condition five, significant between-test differences were found for all

three pairs of tests ($p \leq 0.02$). For SOT condition six, significant between-test differences were found only between pretests 1 and 2 vs. the posttest ($p \leq 0.003$) but not between pretests 1 and 2.

3.2.4 Were the changes in the clinical tests following the training phase related to the changes in lower extremity strength?

To determine whether the unexpected but significant increase in lower extremity strength scores could explain the observed changes in the SOT, LOS, and BBS scores, two repeated measures MANCOVA analysis were run. In the first, these three outcome measures were the dependent variables and the strength scores were the covariate. In the second, the MANCOVA was run on the differences in scores between tests with the difference in strength scores as the covariate.

For both analyses, all previously reported significant between-test effects were maintained. Significant between-test differences were again observed for all three dependent variables following training (pretest 2 vs. post-test, $p \leq 0.004$). The significant between-test differences in the SOT, LOS, and BBS scores cannot be attributed solely to the observed increase in lower extremity strength scores.

4. Discussion

The questions addressed herein were: Do sensory challenge balance exercises, in near isolation from other forms of balance exercise, (1) affect the adaptive process of multi-sensory re-weighting, and (2) lead to improvements in measures of balance and fall-risk? We found evidence that multi-sensory re-weighting was intact before and after

training, shown by a clear adaptive pattern of gain change across conditions. The primary change post-training was not in this pattern of adaptive gain change, but instead appeared to be a decrease in the overall levels of postural responses to dynamic visual and somatosensory stimuli that reflect an improved ability to discriminate and dissociate self-motion from environmental motion, as well as improvements in three clinical measures of balance.

The difference in touch gains post-training reached statistical significance, but the difference in vision gains did not. This failure to reach statistical significance may be due to the larger variance in the vision versus touch gains which reduced the power to detect between-test differences in vision gain. The absolute level of reduction in vision gains was larger, however, than the reduction in touch gains. The post-training decline in overall vision and touch gains cannot be attributed to an overall reduction in residual postural sway, since there were no significant differences between test periods in either position or velocity variability.

The U-shaped velocity variability profiles illustrate a trade-off between the degree of weighting to a stimulus versus the precision of estimating body dynamics. Large amplitude sensory inputs are ‘down-weighted’ to minimize responses that would threaten stability if, for example, coupling to vision remained high. However, the consequence of down-weighting vision is reduced sensory information available for estimation, leading to increased sway variability. Conversely, it is advantageous to ‘up-weight’ small amplitude sensory inputs because even strong coupling does not threaten stability. Up-weighting means more information is available for estimation, hence a reduction in sway variability. Moreover, inter-sensory re-weighting is a compensatory mechanism for

down-weighting a particular stimulus. When vision is down-weighted, touch is up-weighted (and vice versa). However, the increase in velocity variability as stimulus amplitudes increase indicates that inter-sensory re-weighting cannot compensate completely for information loss due to down-weighting.

The current finding [that sensory challenge balance exercises lead to lower overall levels of vision and touch gain] is consistent with prior intervention studies that used the SOT or ‘Clinical Test of Sensory Interaction on Balance’ to reflect multi-sensory re-weighting capabilities (Anacker and Di Fabio, 1992;Hu and Woollacott, 1994;Rose and Clark, 2000). Experience and practice may improve perception of sensory conditions in the environment, and estimation of body position and motion within the environment, or both.

Contrary to previous research suggesting deficiencies in sensory re-weighting in the elderly (Stelmach, Teasdale et al., 1989;Stelmach and Worringham, 1985;Teasdale, Stelmach et al., 1991b;Teasdale, Stelmach et al., 1991a;Whipple R, 1989), the present and previous findings in our laboratory (Allison et al., 2006, in press) have shown that multi-sensory re-weighting is intact in fall-prone older adults with sufficiently intact peripheral sensation. Recent results provide further support that the elderly are not deficient in the re-weighting process per se, but in the discrimination of stimulus amplitude and time scale over which they re-weight (Allison, Kiemel et al., 2005).

Clinical measures of balance and fall-risk

SOT, LOS and BBS: The pre-training findings of lower SOT scores, restricted LOS and reduced BBS scores was entirely consistent with results from previous studies

(King, Judge et al., 1994; Wallmann, 2001; Hu and Woollacott, 1994; Rose and Clark, 2000; Whipple, 1997). After training, we observed increases in: 1) SOT scores in the three conditions in which the surface was moving; 2) LOS scores; and 3) BBS scores.

The SOT findings are particularly relevant to the experimental findings. Because the SOT ‘forces’ the postural control system to shift it’s reliance from one sensory information source to another, it is typically thought to reflect sensory re-weighting capabilities. Previous studies reporting SOT results for populations of healthy and fall prone adults found that SOT scores are lower in healthy older adults and especially so in the fall-prone elderly (Woollacott et al, 1986; Horak, Shupert et al., 1989; Wolfson, Whipple et al, 1992; Whipple, Wolfson et al, 1993; Baloh, Fife et al, 1994; Baloh, Spain et al, 1995; Baloh, Corona, 1998). For both groups, the largest deficiencies are seen in sensory conditions where both visual and somatosensory inputs are simultaneously altered (conditions five and six). Our pre-test results were consistent with prior research.

The greatest post-training increase in sway control was seen in the three conditions in which the surface was moving (SOT conditions four, five and six). Improved sway control in condition four may be interpreted as an improved use of visual inputs, and improved sway control in conditions five and six may be interpreted as improved use of vestibular inputs.

Unexpected clinical findings

Activities-specific Balance Confidence scale (ABC): We expected, but did not find, a post-training increase in the ABC scores. Previous studies have shown a strong correlation between ABC scores and actual balance performance levels (Cho, Scarpace et

al., 2004; Hatch, Gill-Body et al., 2003; Kressig, Wolf et al., 2001). As post-training improvements in balance were expected, we posited that the ABC scores would also improve. In fact, a subset of subjects demonstrated a rise in ABC scores, but scores in a second subset of subjects were reduced. This led to an overall lack of significant change for the group as a whole.

Lower extremity strength: The observed increase in strength scores was unexpected, especially in light of prior research demonstrating that balance exercises alone do not produce an increase in strength (Wolfson, Whipple, Derby, et al, 1996). Because the goal of the sensory challenge exercise program was to improve balance via improvements in multi-sensory integration and re-weighting, resistance and flexibility exercises designed to increase strength and joint range of motion, respectively, were intentionally excluded. We consider two explanatory possibilities.

First, the observed strength increase was real, and resulted merely from having a group of previously sedentary, weak older adults transit to the therapy department and perform standing exercises for 45 minutes twice a week. Our results may differ from those of Wolfson et al because of differences in initial subject strength, the force demands of the exercise program, and the method of strength measurement.

A second possibility is that the observed strength increase was not real in the sense that no true muscle hypertrophy occurred. It is well-documented that resistance exercise of sufficient intensity and duration is required to produce significant increases in muscle strength in older adults (Hakkinen et al, 2001; Brandon et al, 2003; Fiatarone et al, 1990). Since the subjects in our study did no formal resistance training at all, it is difficult to attribute the change in their strength scores to actual changes in muscle. An

alternative explanation related to the failure of central activation in older adults has been previously proposed (Stackhouse, Stevens et al., 2001; Stevens, Stackhouse et al., 2003).

The unexpected finding of increased strength raised the possibility that we might not be able to isolate the primary possible cause of any changes in the clinical balance measures to multi-sensory integration and re-weighting mechanisms. However, the two ANCOVA analyses revealed that the increases in strength could not account entirely for the increases in SOT, LOS or BBS scores. We therefore discount the possibility that improvement in strength was the primary reason for improvements in these three clinical balance measures.

Changes in SOT and BBS scores during the control phase: We hypothesized that there would be no significant change in any of the laboratory or clinical measures during the control phase. However, we observed significant increases in the SOT and BBS scores between PreTest 1 and PreTest 2. This finding hampers a straightforward interpretation of the positive effects of the training program on SOT and BBS scores, as we saw increases in these scores both with and without training.

One possible explanation is measurement error, implying that the SOT and BBS have poor test-retest reliability. However, prior research has shown that the SOT has moderate test-retest reliability (Ford-Smith, Wyman et al., 1995), and that this reliability improves if subjects perform the SOT twice at each testing session and scores from the second test are used. Further, the BBS has very high test-retest reliability (Berg, Wood-Dauphinee, and Williams, 1995).

A second explanation is that the subjects' performance at PreTest 2 was truly different from their performance at PreTest 1 because the physical condition and

performance level of study participants were fluctuant. Two findings argue against this rationale. First, we would expect that fluctuations across the group would be random. Instead, almost all of the change from PreTest 1 to Pretest 2 was in the positive direction. Second, we found no significant differences during the control phase in the laboratory measures or scores on the ABC Scale, LOS test, strength and range-of-motion. These results argue for stable, not fluctuant, control phase performance.

Summary and conclusions

In this study we used a specific, quantitative measure of multi-sensory re-weighting to demonstrate that sensitivity to dynamic environmental stimuli in the fall-prone elderly is altered following a sensory-challenge balance exercise program. The range of gain scores in these fall-prone older adults was shifted lower to more closely match the range of gain scores seen in healthy young adults; we interpret this change as an improvement in the ability to discriminate and dissociate self-motion from environmental motion. Concurrently we found improvements in three clinical measures of balance, one of which reflects a reduction in fall risk after the training program [BBS] (Shumway-Cook et al, 1997; Lajoie & Gallagher, 2004). These results offer initial support for the hypothesis that multi-sensory integration may be one of the mechanisms through which sensory challenge balance exercises effect improvements in balance, and improvements in this mechanism may generalize to other components of postural control. Further work is required to demonstrate a causal relationship, and to determine the relative contribution of sensory challenge balance exercises when used in conjunction

with other forms of balance exercise.

We propose that investigations of interventions on balance and fall-risk ought to include (1) hypotheses about postural control mechanisms related to the intervention, and (2) specific measures of the postural control mechanisms in addition to behavioral outcome measures of balance. Such knowledge could be used to more specifically customize the prescription and provision of balance exercise programs for fall-prone older adults, and thereby improve outcomes.

Figure Captions

FIGURE 3-1

Experimental set-up.

A subject is illustrated in a modified tandem stance facing the computer generated visual display while lightly contacting the touch surface with her right fingertip. Markers on the occiput and lower lumbar region tracked the trajectories of the head and estimated center of mass, respectively.

FIGURE 3-2 A – C

Time series.

Three exemplar time series are shown from one fall-prone older adult at (top) Pre-test 1, (center) Pre-test 2, and (bottom) Post-test. A 125-s segment of data from each test session shows the time series of visual display motion at 0.2 Hz (upper trace), medio-lateral center of mass displacement (middle trace), and touch surface motion at 0.28 Hz (lower trace). In these exemplars, visual motion amplitude is 4mm and touch surface motion amplitude is 2mm.

FIGURE 3-3 A – E

Amplitude spectra.

Five exemplar amplitude spectra are shown for (A) visual display motion, (B) touch surface motion, and center of mass displacements from the same fall-prone older adult in Figure 2 at (C) Pre-test 1, (D) Pre-test 2, and (E) Post-test. In these exemplars, visual motion amplitude is 4mm and touch surface motion amplitude is 2mm. Note the lower Post-test COM response amplitudes to the dynamic vision and touch stimuli.

FIGURE 3-4 A – D

Group mean gain and phase results.

The mean (\pm S.E.) (A) visual gain, (B) touch gain, (C) visual phase, (D) touch phase versus condition at each test period.

FIGURE 3-5 A – B

Group mean position and velocity variability results.

The mean (\pm S.E.) (A) position variability and (B) velocity variability versus condition at each test period.

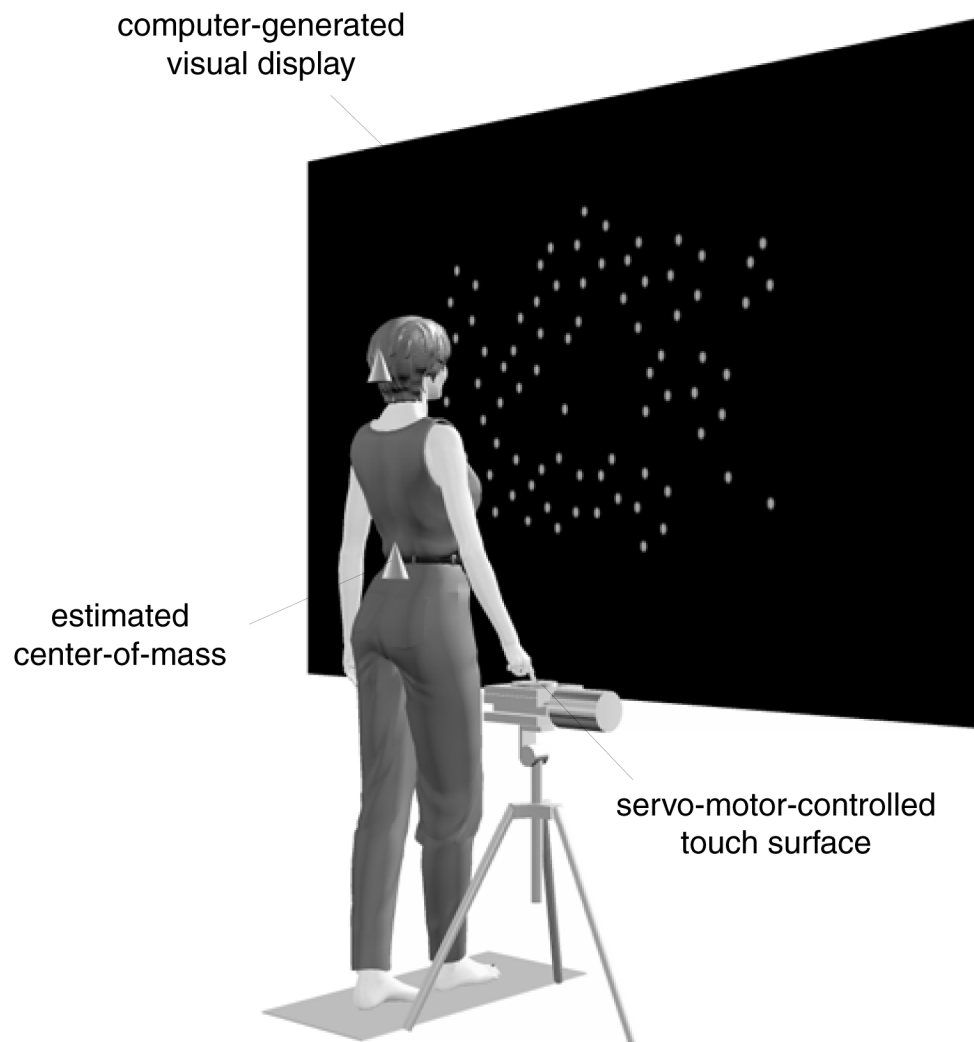


Figure 3-1: Two-frequency vision and touch experimental paradigm.

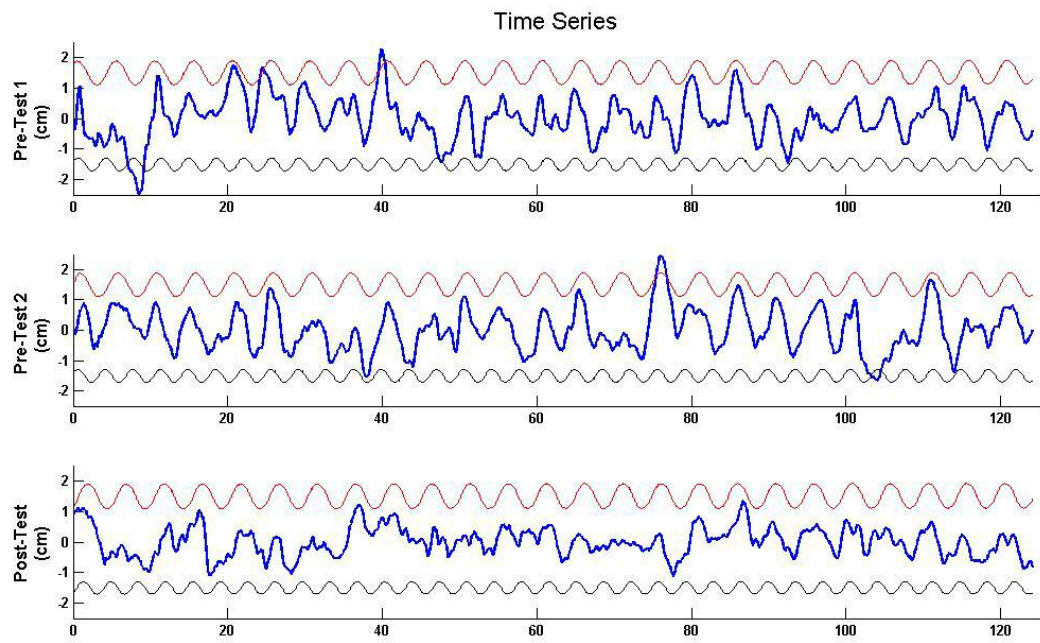


Figure 3-2: Exemplar time series data.

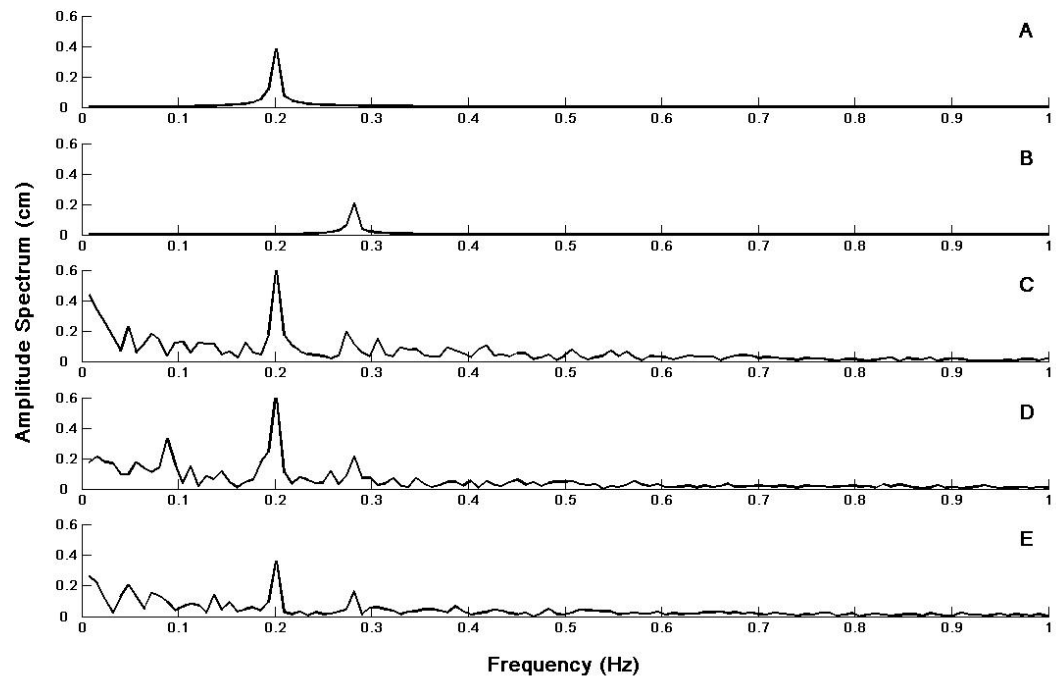
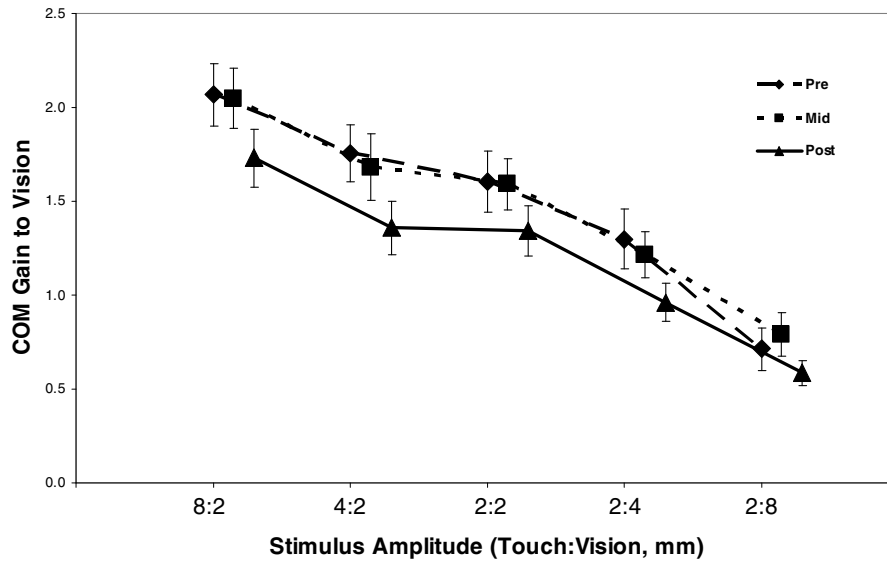


Figure 3-3 A -E: Exemplar amplitude spectra data.

Training Effect: Vision Gain



Training Effect: Touch Gain

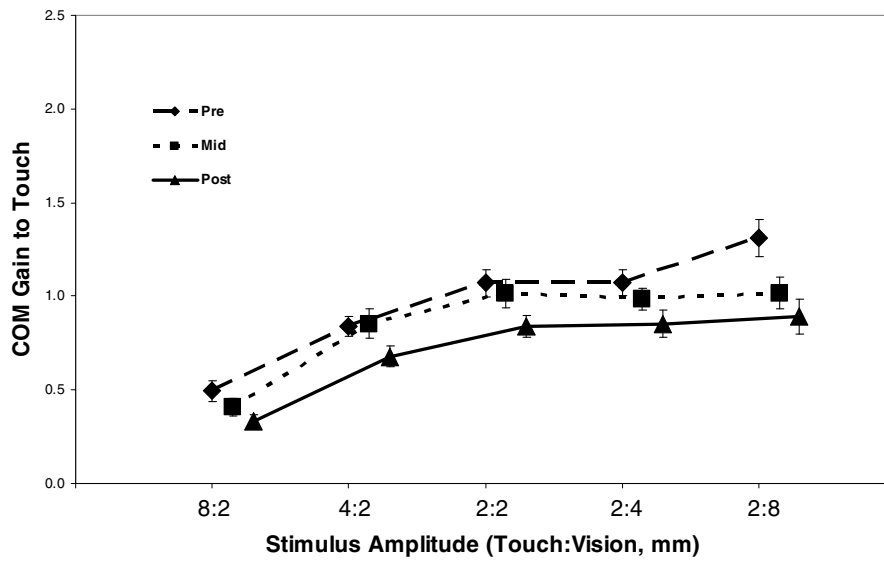


Figure 3-4 A and B:
Effect of Training on Group Vision and Touch Gain

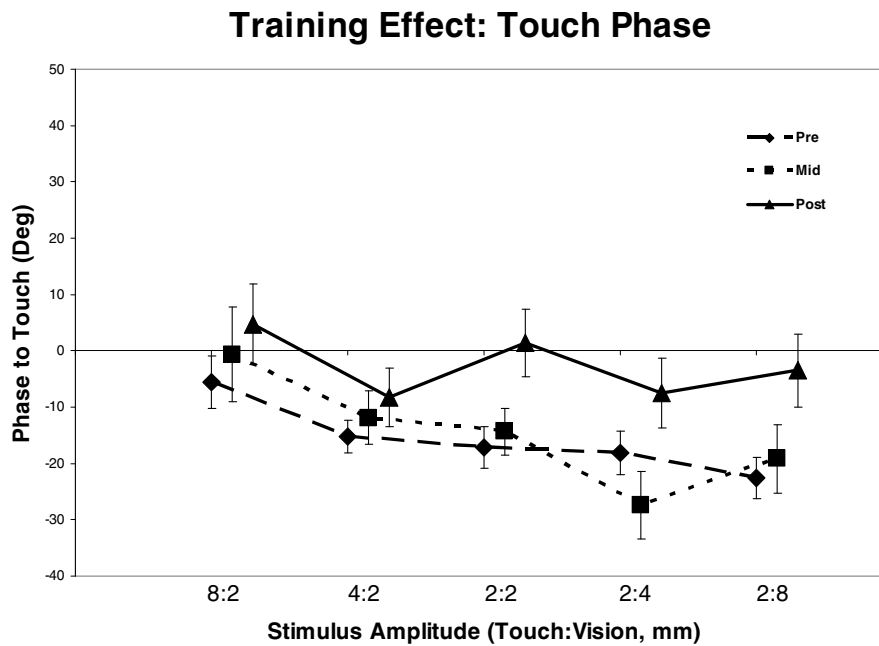
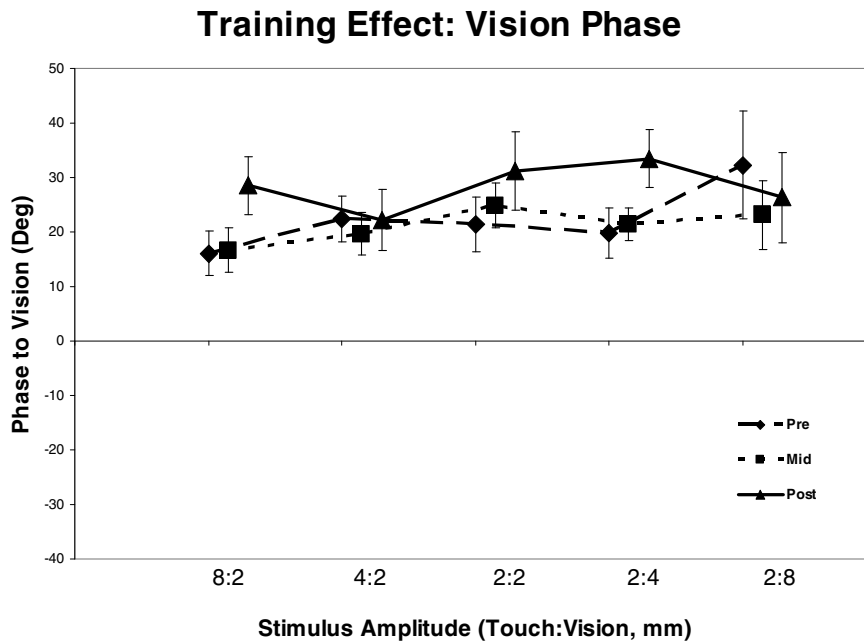
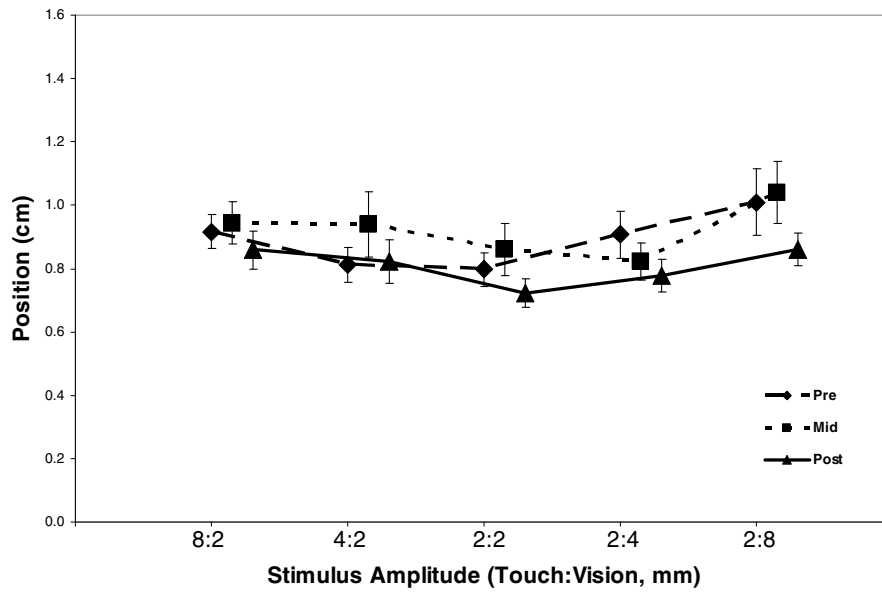


Figure 3-5 A and B:
Effect of Training on Group Vision and Touch Phase

Training Effect: Position Variability



Training Effect: Velocity Variability

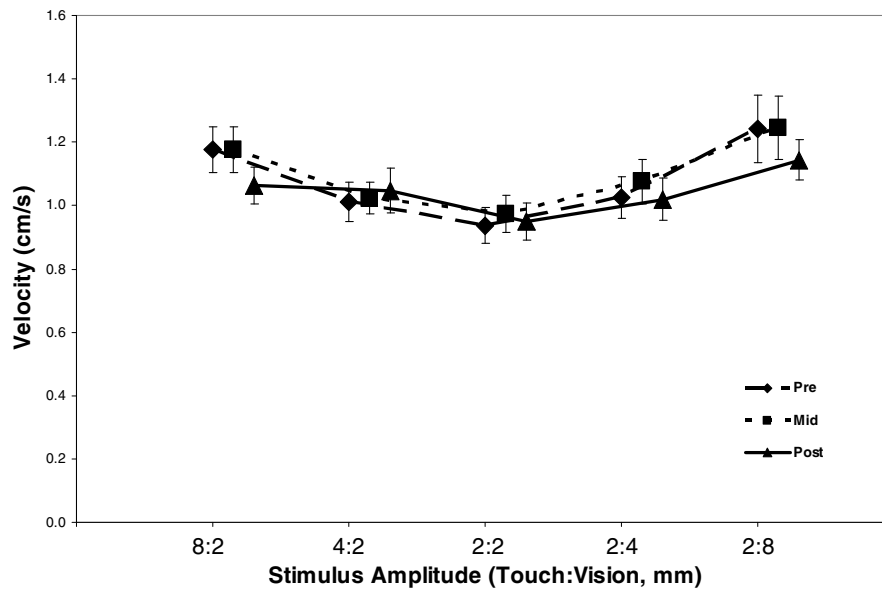


Figure 3-6 A & B:
Effect of Training on Group Position and Velocity Variability

Chapter 4:

The Dynamics of Multi-sensory Re-weighting in Healthy and Fall-prone Older Adults

1. Introduction

Falls in the elderly are dangerous, debilitating and costly (Englander, Hodson et al., 1996; Tinetti, Baker et al, 1994; Tinetti and Williams, 1998). Much research is needed to identify the causative mechanisms responsible for imbalance and falls in older adults. Impairments in multi-sensory integration and re-weighting contribute to impaired balance control (Horak, Shupert & Mirka, 1989; Alexander, 1994; Teasdale et al, 1991a, 1991b; Hay et al, 1996; Woollacott, 2000). Specifically, Teasdale et al (1991b) have stated that ‘an increased slowness in the processing of information from vestibular, visual, and somatosensory systems could contribute greatly to a decline in postural stability’ in older adults. Peterka & Loughlin (2004) indicate that a slower time course of multi-sensory re-weighting in older adults would mean a lengthened period of higher postural instability and a probable increased risk for falls. The purpose of this study was to characterize differences in the time required to adaptively reweight visual information in healthy young, healthy older, and fall-prone older adults. Enhanced understanding of age and impairment related changes in balance control mechanisms may lead to improvements in the care of older adults with instability and falls.

Vestibular, visual and somatosensory information detected peripherally is subsequently integrated in the central nervous system (Horak & Macpherson, 1996). This integrated information is used to produce estimations of body position and motion in

space that are constantly updated as we move through the environment. As sensory conditions change, e.g. moving from a well-lit to a dark room, or a fixed to a less stable surface, the postural control system identifies and selects the sensory inputs that provide the most salient, reliable information (Horak & Macpherson, 1996). This process is called ‘multi-sensory re-weighting’ [MSR].

The ability to select and reweight alternative orientation references adaptively is one of the most critical factors for postural control in elderly populations (Horak et al, 1989). Central sensory selection and re-weighting processes degrade with increasing age and are particularly deficient in fall-prone versus healthy adults (Horak et al, 1989; Alexander, 1994; Teasdale et al, 1991b; Hay et al, 1996; Woollacott, 2000). Previous research indicates that MSR deficits are in part responsible for the postural control problems seen in older adults with a history or high-risk of falls (Stelmach et al, 1989; Teasdale et al, 1991a, 1991b, 1993; Hay et al, 1996). For all ages, postural instability increases when sensory information is withdrawn, as the acquisition of sensory information necessary to perceive body position/motion is compromised. Strong support for the existence of central MSR deficits in healthy and fall prone older adults comes from studies that demonstrate increased postural instability not just when sensory information is withdrawn but even when accurate visual and/or somatosensory information is re-introduced or added (Teasdale et al, 1991b; Hay et al, 1996).

Though results from these previous studies offer support for deficient MSR processes in healthy and fall prone older adults, none were designed to specifically investigate the time-course of MSR. Importantly, many earlier studies used postural sway measures such as mean sway amplitude that by themselves cannot determine whether

changes in sway are due to sensory processes, motor processes, or some combination of both (c.f., Anacker and Di Fabio, 1992; Colledge, Cantley, Peaston, Brash, Lewis, and Wilson, 1994; Whipple R, 1989). The use of these measures in sensory studies is problematic because results might be affected by deficient motor processes, e.g. weakness, slow reaction times, etc. Further, mean sway measures are a gross measure of postural performance that do not identify specific mechanisms actually used by the postural control system. The current study used improved measures that allowed us to quantify changes in the coupling of postural sway to visual inputs when visual motion conditions change.

Jeka, Kiemel and Oie developed a multi-sensory experimental paradigm that simultaneously manipulates two sensory inputs (vision and touch) to investigate MSR. Using this protocol, Oie, Kiemel & Jeka (2002) quantitatively demonstrated MSR in a group of healthy young adults. Subsequently, Allison et al (in press) applied this same paradigm to explore MSR in healthy and fall-prone older adults. Evidence for MSR was found in all three groups; the pattern of adaptive change across sensory conditions was strikingly similar. This finding was surprising and inconsistent with results from prior studies that implicated deficient central MSR processes as a cause of instability and fall risk in older adults.

One explanation for the discrepancy in findings is that prior studies used relatively short duration trials (typically 10 to 30 seconds) while Allison et al (in press) employed longer duration trials (2 minutes). Slowed central sensory processing is well-documented in older adults. Healthy and fall-prone older adults may be able to reweight visual and somatosensory information, but perhaps not as quickly as young adults. If this

is the case, then differences in MSR between young, healthy older, and fall-prone older adults may be more apparent when measured over a short time period and relatively obscured when measured over a long time period. Figure 4-1 illustrates this point. The postural sway of one fall-prone older adult is shown over a 60 second time period, with initially large amplitude sway that gradually dampens over time.

Does MSR occur more slowly in older and fall-prone adults versus young adults? This is an important question because postural stability is relatively compromised until re-weighting is completed. A slower time course of MSR in older and fall-prone adults would mean an extended period of heightened postural instability and a probable increased risk for falls (Peterka & Loughlin, 2004).

In this experiment we investigated the time-course of MSR by changing the amplitudes of visual motion stimuli within a trial, versus between trials as in previous studies. We hypothesized that MSR is a relatively slow process (on the order of seconds), that MSR occurs more slowly in older versus younger individuals, and most slowly in fall-prone adults. Such a finding would have implications for the assessment and treatment of fall-prone older adults, since a lengthened duration of MSR produces a longer period of instability and potentially increases the risk for falls when sensory conditions change. Our previous finding that MSR may be impaired but is not absent in fall-prone adults (Allison, et al, in press) offers the possibility that MSR processes may respond to therapeutic interventions to reduce instability related to environmental motion. Indeed, results from our subsequent intervention study support this notion (Allison et al, submitted). Therapeutic interventions for the fall-prone elderly can become more targeted

and effective if we achieve an improved understanding of the mechanisms that lead to postural instability.

2. Methods

2.1 Subjects

Three groups of subjects were recruited, healthy young adults (n=21; mean age 23 yrs, range 20-30 yrs), healthy older adults (n=25; mean age 76.6 yrs, range 70-93) yrs, and fall-prone older adults (n=17; mean age 79.9 yrs, range 72-92) yrs. All subjects were interviewed and informed of the experimental procedures approved by the Institutional Review Board. Because groups of fall-prone older adults tend to have larger numbers of women, we attempted to match the ratio of women to men in the healthy older and healthy young adult groups to the gender proportions in the fall-prone older adult group. The ratio of females to males in the fall-prone, healthy older and healthy young groups was 60/40%, 56/44%, and 66/34% respectively.

Healthy young subjects (n = 21) were drawn from volunteers recruited from the staff and student population at the University of Maryland and the surrounding greater Washington, DC – Baltimore area. Young subjects met the following inclusion criteria: 1) no history or evidence of orthopedic, muscular, psychological or neurological disability, 2) normal strength and joint range of motion, 3) no current medication that might affect balance, and 4) vision corrected to 20/20.

Older adult subjects, age 70 years and older, with and without a history/high risk of falls were recruited from the university Alumni Emeritus Club, nearby retirement

centers, and the local community. Individuals age 70 years or older were included, as one-in-three people this age range are prone to falls. Older subjects were recruited by posted flyers and local seniors' newsletters and papers. All elderly subjects met the following inclusion criteria: 1) no history or evidence of psychological or neurological disability, 2) no evidence of current or recent acute illness, 3) no evidence of current, recent or changing orthopedic disability, 4) no clinically detectable peripheral somatosensory or vestibular loss, 5) no current medication that might affect balance, 6) limited polypharmacy, 7) vision sufficient to read, watch TV and drive, 8) ability to stand for three minutes at a time without upper extremity support, 9) ability to tolerate a three hour testing session.

Subjects in the healthy older adults group (n = 25) had no history of falls and no reported symptoms of imbalance. Further, these healthy older adults reported a high level of daily activity and some form of regular, moderate intensity exercise or recreational activity. Subjects in the fall-prone older adults group (n= 17) had a history of falls or were at high risk for falls as determined by 1) reported incidence of multiple near falls, 2) balance decline with reduction in functional activity level due to imbalance, and 3) scores on the Berg Balance Scale and/or the Timed-Up-and-Go Test. Older adults with and without a history or high risk of falls were included to permit comparison of MSR processes between these two groups. There were no significant differences between the two older groups in age (mean = 76 for HO and 79 for FP, range = 70-93 for HO and 72-92 for FP), number of medications taken (mean = 2 for HO and 3 for FP, range 0-6 for both groups), or scores on the Mini-Mental Status Exam (mean = 29.5 and range 28-30 for both groups).

2.2 Experimental Set-Up

Apparatus: The visual display system was a Fakespace Systems CAVE[®] Automatic Virtual Environment, a room-sized advanced visualization tool that combines high-resolution, stereoscopic projection and 3-D computer graphics to create a virtual environment. The CAVE is a rear projected 8' x 10' three wall display system with seamless corner technology that produces a sense of spatial immersion. Body kinematics were measured using an Optotrak system (Northern Digital, Inc) to calculate the body's estimated center of mass (COM) position from a summation of individual segment COMs (Kane & Levinson, 1985). Force transducers embedded in the support surface measured vertical reaction forces and allow for calculation of center of pressure (COP) displacement under the foot. A PC workstation along with LabView data acquisition software were used to simultaneously generate visual displays and collect data from the force platform and Optotrak markers.

2.3 Experimental Protocol

2.3.1 Clinical Screening

All older adult subjects underwent a battery of clinical tests as listed below to determine eligibility for continued laboratory testing, which occurred on a separate day. All clinical tests were administered by one licensed physical therapist. The initial testing session, including orientation, explanation, informed consent procedures and clinical screening tests took one and one-half to two hours to complete. Subjects performed the following clinical testing procedures:

Folstein Mini-Mental Status Exam, a short verbal and written test of orientation and memory.

Visual Acuity using a standard Snellen eye chart.

Dynamic Visual Acuity Test, a screening test for the vestibular-ocular reflex (VOR) that involves reading the same Snellen eye chart while the head is being passively rotated by the examiner at a frequency of 2-3 Hz.

Halmyagi “Head Thrust” Maneuver, a screening test for bilateral peripheral vestibular loss that involves voluntary visual fixation during unexpected rapid passive head movement.

One Minute Sit-to-Stand Test, a screening test for bilateral lower extremity strength and endurance that requires subjects to stand up as many times as possible in 60 sec.

Bilateral Lower Extremity Somatosensory Testing. *Touch* was evaluated using Semmes-Winstein monofilaments on the sole of the foot. *Vibration* was evaluated using a 128 Hz tuning fork at the first metatarsals and medial malleoli. *Proprioception* was evaluated using passive motion at the great toe and ankle joints.

Berg Balance Scale. This balance test consists of 14 activities ranging in difficulty level from low (rising from a chair) to high (standing on one foot). This test is highly reliable and provides an estimation of fall risk.

Timed Get-Up-and-Go Test. This timed mobility test requires the seated subject to rise, walk three meters, turn around, walk back, and sit down. This test is highly reliable and provides an estimation of fall risk.

2.3.2 Laboratory multi-sensory re-weighting Protocol

Laboratory testing occurred on a day separate from and subsequent to the clinical testing. The informed consent form and laboratory testing procedures were reviewed. The MSR experiment took two to three hours to complete. A trained research assistant (RA) stayed with each older subject at all times and was responsible for the comfort and safety of the older subjects

Subjects stood with standardized foot positioning based on height (McIlroy & Maki, 1997) in the center of the stationary force platform. Subjects faced a three-sided projection screen enclosure that displayed a random pattern of 500 small triangles; no triangles were located directly on the area of the front screen corresponding to the subject's eye height (Figure 4-2).

Subjects wore a harness secured to the ceiling by two straps. The straps were adjustable to allow subjects to sway without impedance; thus subjects were not prevented from swaying or losing their balance but were prevented from falling. On each subject were placed seven markers at each of the following body landmarks: ankle, knee, hip, shoulder, head left, head right, and head top. Data from the four lower markers are used to calculate the subjects' COM trajectory in response to the visual motion stimuli.

Subjects were instructed to stand as steadily as possible without stiffening. The subjects began each trial by looking straight ahead at the blank area of the anterior visual display screen. Trials were 210 seconds in duration including 180 seconds of visual motion sandwiched between static visual display at the beginning and end of each trial. Subjects were thus unable to predict when visual motion might begin. After each trial,

subject sat comfortably for at least two minutes. Subjects were instructed to stop the trial immediately at any time if they felt faint, dizzy, fatigued, uncomfortable, etc.

The visual display oscillated at a constant frequency of 0.4 Hz. initially at one of two different amplitudes: 3 mm or 12 mm. The 3mm stimulus motion was not readily perceptible while the 12 mm stimulus motion was clearly detectable. After 60 seconds the visual display switched from 3 mm to 12 mm or vice versa, and remained at this amplitude for 120 seconds (Please see Figure 4-3). A longer pre-switch time segment would have been ideal, however, the limited tolerance of the fall-prone older adults was a primary consideration in setting the total trial length at three minutes. The order of the visual display amplitude switches (high-to-low vs. low-to-high) was randomized across subjects.

A total of 11 trials were performed. The first trial was a calibration trial under lighted conditions with no visual motion and voluntary subject weight shifts to maximal excursions in the anterior, posterior, left and right directions. The room was then darkened for several minutes to permit adjustment to the dark. Subsequently 10 visual motion experimental trials were performed. During the testing trials the room was silent, but conversation was encouraged during the rest periods.

2.4 Analysis

The anterior-posterior COM trajectories were calculated according to Winter's method (Winter, 1990, 1991). Figure 4-4 A-C shows exemplar time series of the anterior-posterior COM and the visual motion stimulus. The amplitude spectrum was computed from the time series by taking the absolute value of the Fourier transform (Fig. 4-5 A-C).

The frequency response function (FRF) at the stimulus frequency (0.4 Hz) was obtained by dividing the transform of the COM by the transform of the visual stimulus, resulting in a complex-valued FRF.

Gain and phase: From the frequency spectra, we obtained values for gain and phase. Gain values represent the ratio of the COM response amplitude at the vision stimulus frequency (0.4 Hz) to the vision stimulus amplitude. Gain is calculated as the absolute value of the FRF at the stimulus frequency. For example, if the visual stimulus amplitude at 0.40 Hz is 0.3 cm, and the COM response amplitude at 0.40 Hz is 0.6 cm, then the vision gain value for that trial is $0.6/0.3 = 2$. A change in gain to a changing visual motion stimulus would represent adaptive intra-modality re-weighting. We interpret higher versus lower vision gains to be indicative of higher weighting of, and stronger coupling to, the visual stimulus.

Phase values, which represent the temporal relationship between the visual motion stimuli trajectory and the COM response trajectory, were computed as the argument of the transfer function and converted to degrees. Positive phase values indicate that the COM trajectory leads the visual stimulus trajectory, while negative phase values indicate that the COM trajectory lags the visual stimulus trajectory.

Position and velocity variability: Position and velocity variability of postural sway are the standard deviation of the residual COM displacement and its derivative, respectively, after removal of the postural sway response at the stimulus frequencies (cf. Jeka et al., 2000). The residual COM displacement was computed by subtracting sinusoids corresponding to the COM Fourier transform at the touch and vision stimulus frequencies. The derivative of the residual COM trajectory was computed by finite

differences using every fifth value of the trajectory, which corresponds to a time step of 0.1 s.

2.4.1 Statistical analysis

Statistical analyses were performed using SPSS[®] version 13 and a non-linear statistical model. There were four measures of interest: gain, phase, position variability, and velocity variability. Two of these (gain and phase) are non-linear functions of the real and imaginary parts of the FRF. To ensure the legitimacy of analysis of the variables derived from the complex-valued FRFs, a preliminary statistical analysis of the real and imaginary parts of the FRF was performed. We divided each condition (High-to-Low and Low-to-High) into two parts (pre-switch and post-switch) to obtain four conditions: Pre-High and Post-Low (from condition High-to-Low) and Pre-Low and Post-High (from condition Low-to-High). A 3x4 (Group by Condition) repeated measures MANOVA was performed using the real and imaginary parts of the complex-valued FRF. Subsequently, gain and phase were analyzed using a non-linear statistical model, while position variability and velocity variability were analyzed using SPSS[®] repeated measures MANOVA.

Gain and phase: Some subjects demonstrated low gains in some time intervals, raising the possibility that small errors in estimating the FRF might lead to large errors in the estimation of phase and a positive bias in the estimation of gain. To minimize this possibility, we did not average gain and phase across trials or subjects but instead averaged the FRF across trials and then across subjects. Subsequently, we used the

absolute value and the argument of the mean FRF as estimates of “group gain” and “group phase”, respectively.

To explore the effect of time on the postural sway response to the visual motion stimulus, we divided each trial into 36 five-second time intervals (two stimulus cycles per interval). The five-second time interval was chosen as a compromise between accuracy and temporal resolution. There were 12 time intervals before the visual motion stimulus amplitude switch and 24 after the switch. The FRFs were computed for these five-second intervals and averaged across the five trials in each condition (High-to-Low and Low-to-High) for each subject, then across subjects in each of the three groups.

For analysis, we specified 12 five-second time intervals, six in each condition, as follows:

Time Interval (sec)	High-to-Low	Low-to-High
0-5	Pre-High Beg	Pre-Low Beg
15-20	Pre-High Mid	Pre-Low Mid
55-60	Pre-High End	Pre-Low End
60-65	Post-Low Beg	Post-High Beg
75-80	Post-Low Mid	Post-High Mid
175-180	Post-Low End	Post-High End

To test for main group and time effects, and a group-by-time interaction, we performed overall tests on the underlying measures (the real and imaginary parts of the FRF). Subsequently, univariate tests for gain and phase were performed, followed by planned post-hoc pair-wise tests.

To capture the rapid changes in gain associated with the switch in visual motion stimulus amplitude, we used FRF values from the last five-second time interval before the switch and the first five-second time interval after the switch (i.e., Pre-High End vs. Post-Low Beg, and Pre-Low End vs. Post-High Beg). To describe the slow changes in gain over time at each amplitude, we fit the FRFs by a linear function of the time-interval index over two separate time periods: between 15-60 sec before the switch (Pre-High and Pre-Low), and between 65-180 sec after the switch (Post-Low and Post-High). The FRFs were fit using least-squares fits of their real and imaginary parts.

Null hypothesis for group gains and phases were tested with the assumption of multivariate normality for the real and imaginary parts of the FRFs. Tests of gain and phase required a non-linear analysis as they are non-linear functions of the FRF. To test any null hypothesis, we first computed the maximum likelihood for the full model and for a model obeying the null hypothesis. The ratio of these likelihoods was used to compute Wilks' Lambda, which was tested using the same degrees of freedom as in the corresponding linear case.

For all three groups together, we tested for group and time main effects and a group by time interaction. Subsequently, we performed these same tests on each pair of groups. The significance levels for these paired group comparisons were protected by the overall univariate test. Planned post-hoc pairwise comparisons then examined the group-

by-time interaction effects. The significance levels for these paired group comparisons were unprotected and are presented below unadjusted.

To determine whether or not there were differences between groups in the absolute levels of gain, we performed group comparisons for each pair of groups at each of the twelve time intervals specified above. To determine between-group differences in the rates of adaptive gain change, we examined how quickly the COM gain values change from one level to another both at the time of the visual stimulus amplitude changes (rapid change) and over the course of the extended exposure to each amplitude (prolonged change). Relatively slower rates of gain change in the healthy and/or fall-prone elderly groups would indicate a prolonged re-weighting process, longer periods of instability, and potentially, a higher risk for falls.

Position variability and velocity variability: We divided each of the two conditions into three 60-second segments, Pre-switch, Post-switch 1, and Post-switch 2. Subsequently we performed a 3x6 (Group by Condition) repeated measures MANOVA with planned post-hoc pairwise comparisons. After determining that there were no significant differences in either position variability or velocity variability between the two 60 second post-switch segments, we collapsed them into a single, 120 second post-switch segment. Then we performed a 3x4 (Group by Condition) repeated measures MANOVA with planned post-hoc pairwise comparisons using Bonferroni correction.

3. Results

3.1 Gain and Phase

We expected that, for all groups, gain will vary inversely with stimulus amplitude. That is, when the visual motion stimulus amplitude is high, gains will be low, and vice versa. Thus, when the stimulus amplitude changes from high to low, we expect that gains will change from low to high, and vice versa. Our results were consistent with this expectation.

Preliminary analysis: The real and imaginary parts of the complex-valued FRFs were explored using a 3x4 (Group by Condition) repeated measures MANOVA. Conditions were: Pre-High and Post-Low (from High-to-Low) and Pre-Low and Post-High (from Low-to-High). The multivariate test revealed a significant effect for Condition ($p < 0.001$), and a significant Group-by-Condition interaction ($p = 0.022$). No significant main effect for Group was observed ($p = 0.093$).

Univariate tests of gain and phase: Gain and phase results for the three groups in the two conditions are shown in Figures 4-6 A-B and 4-7 A-B. Gain and phase were examined using a non-linear statistical model that included the three subject groups and the 12 time intervals described above. The gain analysis revealed a significant main effect for Group ($p = 0.0008$), a significant effect for Time ($p < 0.0001$), and a significant Group-by-Time interaction ($p = 0.045$). Planned post-hoc tests for gain are reported below. The phase analysis revealed no significant main effects for Group or Time, and no significant Group-by-Time interaction (all $p \geq 0.34$). No subsequent analysis of phase data is therefore reported.

Planned post-hoc pairwise tests for absolute levels of gain: Group differences in the absolute levels of gain were observed between both older adult groups and the young adult group, with gains being higher in the older adult groups (fall-prone vs. young, $p=0.0002$; healthy older vs. young, $p=0.0244$). A nearly significant difference was observed between the fall-prone versus healthy older adults, with higher gains in the fall-prone group ($p=0.0562$). (Below we report differences between fall-prone and healthy older adults only when they approach or reach significance.)

To determine differences between groups in the absolute levels of gain at the outset of a change in the visual environment (e.g., at the onset of visual motion), we compared gains between groups at time intervals Pre-High Beg and Pre-Low Beg. (The significance values we report here are unadjusted.) Compared to the young, gains were significantly higher in the fall-prone and healthy older adults when the visual environment first changed from stationary to high amplitude motion (Pre-High Beg, $p<0.0001$ and $p=0.0002$, respectively). When the visual environment first changed from stationary to low amplitude motion, gains were significantly higher in the fall-prone than in the young, and marginally higher in the fall-prone than in the healthy older adults (Pre-Low Beg, $p=0.0048$ and $p=0.0558$, respectively). Gains were not different between the healthy older and the young groups (Pre-Low Beg, $p=0.2217$).

To determine differences between groups in the absolute levels of gain immediately after the visual stimulus amplitude changed, we compared gains between groups at time intervals Post-Low Beg and Post-High Beg. Compared to the young, gains were significantly higher in the fall-prone and healthy older adults when the visual motion amplitude changed from low to high (Post-High Beg, $p<0.0002$ and $p=0.0217$,

respectively). When the visual motion amplitude changed from high to low, however, gains were not significantly different in either the fall-prone or healthy older adult group compared to the young (Post-Low Beg, both $p \geq 0.17$).

To determine differences between groups in the absolute levels of gain at the end of the rapid change in gain, we compared gains between groups at time intervals Pre-High Mid, Pre-Low Mid, Post-High Mid and Post-Low Mid. When the visual motion amplitude was high, gains remained significantly higher in both older adult groups compared to the young (Pre-High Mid, FP vs. Y, $p=0.0021$; HO vs. Y, $p=0.01$. Post-High Mid, FP vs. Y, $p=0.0008$; HO vs. Y, $p=0.0025$). When the visual motion amplitude was low, few differences were seen. Gains in the fall-prone group remained significantly higher than in the healthy older group at time interval Pre-Low Mid ($p=0.0105$), but not at time interval Post-Low Mid ($p=0.2987$). Gains in the fall-prone group also remained significantly higher than in the young group at time interval Pre-Low Mid ($p=0.0002$), but not at time interval Post-Low Mid ($p=0.3728$). Gains in the healthy older group were not significantly different than in the young at either time interval (both $p \geq 0.1475$).

To determine differences between groups in the absolute levels of gain at the end of the prolonged change in gain, we compared gains at time intervals Pre-High End, Pre-Low End, Post-High End and Post-Low End. When the visual motion amplitude was high, compared to young adults, gains remained higher in fall-prone older adults even after prolonged periods of 45 seconds pre-switch (Pre-High End, $p=0.0025$) and 105 seconds post-switch (Post-High End, $p=0.0026$). Compared to young adults, gains in the healthy older adults were marginally higher at the 45 sec. time interval Pre-High End ($p=0.0679$) and significantly higher at the 105 sec. time interval Post-High End

($p=0.0382$). When the visual motion amplitude was low, compared to young adults, gains remained higher in both older adult groups after a 45 second exposure period (Pre-Low End, FP vs. Y, $p=0.0008$; HO vs. Y, $p=0.0174$). However, after a longer exposure of 105 seconds in the low amplitude condition, there were no significant differences in gain between the older adult groups and the young (Post-Low End, both $p\geq 0.1291$).

Planned post-hoc pairwise tests for rates of change in gain: An important difference in the expected directions of gain change with high versus low amplitude stimuli should be noted here. At the onset of High-to-Low trials [transition from stationary to high amplitude], and after the amplitude switch in the Low-to-High trials [transition from low to high amplitude], gains are expected to decrease reflecting that the now unreliable visual stimulus is downweighted. This response to the high amplitude stimulus is the same in Pre-High and Post-High conditions. However, when the stimulus amplitude is low, gain changes are not the same in the Pre-Low versus Post-Low conditions. At the onset of Low-to-High trials [transition from stationary to low amplitude; Pre-Low], gains are expected to *decrease* reflecting that the now less reliable visual stimulus is downweighted. After the amplitude switch in the High-to-Low trials [transition from high to low amplitude; Post-Low], gains are expected to *increase* reflecting that the now more reliable visual stimulus is upweighted. To summarize, gains are expected to decrease in Pre-High, Post-High, and Pre-Low conditions, and to increase in the Post-Low condition. The direction of the gain change depends on whether or not the visual information becomes more versus less reliable.

The gain in each of the three groups showed a highly significant dependence on time (all $p\text{'s} < 0.0001$). Group differences in the rate of change in gain (a group by time

interaction) were not observed between the healthy versus fall-prone older adults ($p=0.4412$), but were observed between both older adult groups and the young adult group (FP vs. Y, $p=0.0480$; HO vs. Y, $p=0.0503$).

To examine the rapid change in gain associated with the switch in visual motion amplitude, we first examined within each group whether there was a significant change in gain following the switch. (The significance values we report here are unadjusted.) For all three groups and both switches (High-to-Low and Low-to-High), we found a significant rapid change in gain (Pre-High End vs. Post-Low Beg, gains increased, all $ps \leq 0.048$; Pre-Low End vs. Post-High Beg, gains decreased, all $ps \leq 0.002$). To determine if there were between-group differences in the rapid change in gain, we compared these changes between groups. For both switches, there were no between-group differences in the rapid rate of gain change (all $ps > 0.28$).

To examine the prolonged change in gain with continued exposure to a single visual motion amplitude, we first examined within each group whether there was a significant change in gain over the course of the extended time period after the rapid change. In the Pre-High and Pre-Low conditions this extended time period is 45 seconds in duration. In the Post-High and Post-Low conditions it is 105 seconds in duration.

In the Pre-High condition, decreases in gain in the fall-prone and healthy older adult groups approached significance, while no significant decrease was observed in the young (FP $p=0.0641$, HO $p=0.0673$, Y $p=0.6040$). Only the fall-prone group demonstrated a significant change in gain in the Pre-Low condition, displaying a surprising continued decrease in gain over time (FP $p=0.0237$, HO $p=0.8009$, Y $p=0.2043$). Prolonged changes in gain were more apparent in the Post-High and Post-

Low conditions, where a longer exposure to the visual motion stimulus occurred. In the Post-High condition, decreases in gain approached significance in the fall-prone group, but were not significant in the healthy older or young groups (FP $p=0.0673$, HO $p=0.1105$, Y $p=0.0812$). In the Post-Low condition, there were significant increases in gain for the healthy older and young groups, and a marginal increase in gain for the fall-prone group (FP $p=0.0590$, HO $p=0.0007$, Y $p=0.0227$).

To determine if there were between-group differences in the rate of prolonged gain change, we compared these changes between the groups that had demonstrated a prolonged change in a given condition. In the Pre-High condition, the young group showed no additional change while the healthy and fall-prone older adults did show a prolonged change. There were no significant differences in the rate of prolonged change between the healthy and fall-prone older adult groups in this condition ($p=0.66$). In the Pre-Low and Post-High conditions, the young and healthy older groups showed no additional change while the fall-prone older adults did show a prolonged change. No between group differences in the rate of prolonged change were analyzed in these conditions as only one group showed such a change. In the Post-Low condition, all three groups showed further change. However, no significant differences in the rate of prolonged change were found between any of the three groups (all $ps \geq 0.35$).

3.2 Position variability and velocity variability

We expect that variability will vary inversely with gain. That is, when gains are high and MSI processes are obtaining reliable information for position and velocity estimation, variability will be low. Conversely, when gains are low and MSI processes obtain less information for estimation, variability will be higher. Based on results from previous studies (Jeka et al, in press; Allison et al, in press), we expected velocity variability, but not position variability, to reflect this relationship.

Position variability and velocity variability results for the three groups in the two conditions are shown in Figures 4-8 A-B and 4-9 A-B. We performed a 3x4 (Group-by-Condition) repeated measures MANOVA. The overall multivariate test revealed significant main effects for Group ($p=0.029$) and Condition ($p<0.001$), and a significant Group by Condition interaction ($p=0.006$). Within-subjects univariate tests for the condition main effect and group-by-condition interaction were not significant for position variability ($p=0.763$ and $p=0.090$, respectively). For velocity variability, a significant condition main effect and a significant group-by-condition interaction were found ($p<0.001$ and $p=0.022$, respectively). Between-subjects univariate tests for the group main effect were not significant for either position variability or velocity variability ($p=0.496$ and $p=0.596$, respectively).

Velocity variability was examined further via planned post-hoc pairwise comparisons with Bonferroni correction. Velocity variability in the Post-Low condition was significantly lower than in all other conditions ($p<0.001$ for all pairs). There were no significant differences in velocity variability between any of the other conditions (Pre-

High, Pre-Low, or Post-High). There were no significant between-group differences in any of the four conditions (all $p \geq 0.19$).

4. Discussion

To explore age- and impairment-related changes in MSR, we compared differences in (1) the absolute levels of gain, and (2) the dynamics of adaptive re-weighting of visual information as reflected by changes in gain over time, in healthy young, healthy older, and fall-prone older adults. When visual motion amplitude was high, both older adult groups had higher absolute levels of gain than the young group. Fall-prone and healthy older adults responded similarly to the stimulus when the amplitude was high. This finding may reflect an age-related change in MSR. Such a change may compromise postural control compared to the young but does not appear to be *by itself* a cause of falls. When visual motion amplitude was low, higher absolute levels of gain were observed in the fall-prone older adult group versus the young and healthy older adult groups. Healthy older and young adults responded similarly to the stimulus when the amplitude was low. This finding may reflect a deficit in MSR in fall-prone elders that cannot be attributed to age alone.

Rapid changes in gain (within 5 seconds) were similar in all three groups for both downweighting and upweighting, indicating little age- or impairment-related deficit in this component of MSR. Examination of prolonged changes in gain, however, demonstrated the compromised MSR processing in older adults, most especially the fall-prone older adults.

Evidence from previous studies has implicated deficient multi-sensory re-weighting as a contributor to instability and falls in older adults (Stelmach et al, 1989; Teasdale et al, 1991a, 1991b, 1993; Hay et al, 1996). Specifically, MSR is thought to be slowed in older adults, both in circumstances where sensory information is withdrawn or becomes unreliable, and when reliable sensory information becomes available after a period of withdrawal (Teasdale et al, 1991a, 1991b, 1993; Hay et al, 1996). Prior studies, however, were not designed to specifically investigate the dynamics of MSR in older adults, nor did they use measures of the postural response that could specifically measure the response to the sensory stimulus separate from other determinants of postural sway. In the current study we exposed participants to a sudden change in visual motion amplitude within each trial and examined the response to this change over a two-minute time period. Measures of gain and phase allowed the response to the visual stimulus to be determined specifically and separately from overall postural sway. Calculation of position and velocity variability permitted the measurement of all residual postural sway not specific to the visual stimulus.

The purpose of this study was to characterize the differences in the time required to reweight visual information in healthy young, healthy older and fall-prone older adults. We observed that, for all groups, following the onset of each trial and amplitude switch there was an immediate rapid change in gain followed in some cases by a continued slow change in gain. To examine these two components of the adaptive response, we divided the trial into 36 five-second intervals and analyzed absolute levels of gain and the relative rate of change in gain at and over selected time intervals. We also observed that absolute

levels of gain were influenced by the amplitude of the visual motion stimulus, with group differences being much more apparent in the high amplitude intervals.

4.1 Group differences in absolute levels of gain

Gain is the ratio of the amplitude of the postural sway response at the stimulus frequency to the amplitude of the stimulus at that unique frequency. In general, gains reflect the degree to which the stimulus has been weighted for central sensory processing; higher gains reflect higher weighting and a stronger coupling to the stimulus. For normal subjects, levels of gain vary with stimulus amplitude (Oie et al, 2002; Allison et al, in press); when stimulus amplitude is high, gains will be low, and vice versa. This variation serves a very functional purpose for the maintenance of postural stability. For example, if postural sway responses remain strongly coupled to the visual scene as its motion increases (i.e., gains remain high), sway will increase until the individual eventually falls. By downweighting a high amplitude stimulus (i.e., gains become low), postural sway is diminished and will remain within the limits of stability. Similarly, upweighting a low amplitude stimulus (i.e., gains become high) permits the individual to use additional sensory information to improve postural control.

A strikingly consistent finding was that differences between both older adult groups and the young group emerged when the visual motion amplitude was high. Gains were higher in both older adult groups than in the young. This result occurred at every selected time interval, both before the switch in the High-Low condition and after the switch in the Low-High condition. The functional implication of this finding is that, for older adults exposed to large amplitude environmental motion, stability may be relatively

compromised. When visual motion amplitude is high, postural control is optimized by reducing the influence of that information source on the MSI estimation process while increasing the influence of other more reliable information sources. In this circumstance, low vision gains would enhance stability whereas high gains would threaten it.

This result is in agreement with earlier studies that have explored ‘visual dependence’ or ‘visual sensitivity’ in older adults using large amplitude visual motion (Sundermeier et al, 1996; Simoneau et al, 1999). Other investigators that have observed a heightened reliance on vision have proposed that it is a compensatory adaptation to an inability to rely on somatosensory and/or vestibular information. This is certainly one possible explanation, but presently there is insufficient evidence to confirm or refute it. To date no studies have simultaneously manipulated stimuli and measured gains in all three sensory systems; until this occurs it will be difficult to say with any certainty whether vision gains in older adults are higher because somatosensory and/or vestibular gains are lower. While it is true that older adults have documented declines in somatosensory and vestibular systems, they also have documented declines in visual systems; it is not immediately obvious that, of the three systems, vision is consistently the best preserved and thus the most reliable.

A second consistent finding is that there were fewer between-group differences when the visual motion amplitude was low. In the three low amplitude time intervals after the switch from High-to-Low, there were no significant differences in the level of gain between any of the three groups. In the first two of the three low amplitude time intervals before the switch from Low-to-High, there were no significant differences in the level of gain between the healthy older adults and the young. Fall-prone older adults did

have higher gains than the young in all three low amplitude time intervals before the switch from Low-to-High, and also had higher gains than the healthy older adults in the first two of those three time intervals. The two primary functional implications of these findings are that (1) solely age-related differences in MSR are not apparent and of no consequence in relatively stationary and reliable visual motion environments, thus MSR does not impair postural stability in healthy older adults in these circumstances, and (2) relative to completely stationary visual environments, even small amounts of motion and instability in the visual environment affect fall-prone older adults and may impair their postural stability.

It is interesting to note that, for healthy older and young adults, levels of gain in the low amplitude time intervals were similar both before the switch in the Low-to-High condition and after the switch in the High-to-Low condition. For fall-prone older adults however, levels of gain were higher in the low amplitude time intervals before the switch in the Low-to-High condition than they were after the switch in the High-to-Low condition. In the Low-to-High condition, subjects transitioned from a completely stationary visual environment to the low amplitude visual motion environment; here, the low amplitude stimulus was relatively *less* stable than the completely stationary environment preceding trial onset. In the High-to-Low condition, subjects transitioned from a highly non-stationary visual environment to the low amplitude visual motion environment; here, the low amplitude stimulus was relatively *more* stable than the highly non-stationary environment preceding the amplitude switch. In all situations where the visual environment became relatively less stationary, whether going from a stationary visual environment to a low amplitude visual motion environment, or from a low

amplitude visual motion environment to a high amplitude visual motion environment, fall-prone older adults had higher gains than healthy older and young adults. They did not downweight vision to the extent that healthy older and young adults did, and this failure to maximally disengage from the non-stationary information source (and presumably to upweight other reliable sources of information) may impair their postural stability.

In summary, (1) fall-prone older adults had higher gains than healthy older and young adults in some low amplitude visual motion time intervals and (2) both healthy and fall-prone older adults had higher gains than young adults in high amplitude visual motion time intervals. Taken together, these findings are in agreement with prior studies of MSR in older adults that used the SOT[®] (Woollacott et al, 1986; Horak et al, 1989; Whipple & Wolfson, 1989; Wolfson et al, 1992; Whipple et al, 1993). In these studies, investigators found that fall-prone older adults were abnormally unstable in any condition where there was a disturbance in the sensory environment, whether the disturbance was mild (SOT conditions 2, 3 and 4) or more severe (SOT conditions 5 and 6). Healthy older adults were not abnormally unstable when disturbances of the sensory environment were mild, but were unstable under the more severely disrupted conditions.

4.2 Group differences in the dynamics of multi-sensory re-weighting

We expected that, for all groups, gains would decrease over time whenever visual motion amplitude became high and when the visual environment changed from stationary to low amplitude motion. We expected that gains would increase over time when visual motion amplitude changed from high to low. Our results were consistent with this expectation. Further, for all groups we observed a rapid change in gain immediately after

the onset of the trial and at the switch in visual motion amplitude, sometimes followed by a more prolonged gradual change in gain over the remainder of the time at a given stimulus amplitude. We examined the rapid change in gain following the sudden change in stimulus amplitude separately from the prolonged change in gain over the remainder of the time segments. We analyzed two periods of prolonged gain change, one that was 45 seconds in duration (from the time before the switch in stimulus amplitude), and one that was 105 seconds in duration (from the time after the switch in stimulus amplitude). For both rapid and prolonged changes in gain, we asked two questions. First, within each group, did any significant change in gain occur? Second, [if so] were there significant between-group differences in the rate of change? We expected that all groups would demonstrate adaptive gain changes, and that these gain changes would be slowest in the fall-prone older adults and fastest in the young adults.

Rapid changes in gain: All groups demonstrated significant changes in gain following both switches in stimulus amplitude, and there were no differences between groups in the rate of this rapid change. We were surprised by the latter finding, as we expected slower adaptation in the healthy older and especially the fall-prone older adults. Such differences might be apparent with a greater temporal resolution than was used here. Even so, this finding has positive implications for healthy and fall-prone older adults. All groups were apparently able to recognize and respond relatively rapidly (within five seconds) to sudden changes in environmental motion. Rapid decreases in gain to suppress unreliable information [when visual motion amplitudes became high] and increases in gain to improve access to reliable information [when visual motion amplitudes became low] are beneficial responses that function to improve postural

stability. Recall, however, that in the Low-to-High condition, absolute levels of gain were higher in healthy and fall-prone older adults both before and after the switch in stimulus amplitude. Despite the similar rapid decrease in gain between groups, healthy and fall-prone older adults still had higher gains after the rapid decrease in gain following the Low-to-High amplitude switch.

Prolonged changes in gain: Following the rapid changes in gain noted above, did further adaptation occur within each group? For young adults, in three of the four time segments (Pre-High 45, Pre-Low 45, and Post-High 105), the answer is no. This implies that young adults accomplished the MSR process by the end of the rapid change in gain. Only during the Post-Low 105 time segment did young adults continue to adapt, as reflected by increasing gains over time. For fall-prone adults, the answer is yes. In all four time segments, gains continued to change over the duration of the time segment, demonstrating continued adaptation. This implies that the MSR process in fall-prone adults is not fully accomplished during the rapid change in gain as further increases or decreases in gain are observed thereafter. Clearly, achievement of full re-weighting takes much longer in fall-prone older adults. Prolonged responses in the healthy older adults were mixed. They demonstrated no prolonged changes in gain during the Pre-Low 45 and Post-High 105 time segments, but did show continued adaptation during the Pre-High 45 and Post-Low 105 time segments.

The only time segment in which all three groups demonstrated continued adaptation was the Post-Low 105 segment. When the visual motion amplitude is low, stability is not threatened and there is little negative consequence to a slower, prolonged upweighting process. However, in time segments Pre-High 45 and Post-High 105, when

the visual motion amplitude is high, stability is threatened. To maximize postural stability, the optimal response here is to downweight vision gains rapidly and completely (assuming that upweighting to other more reliable sources of information also occurs). Only the young consistently behaved in this manner; fall-prone older adults never did.

In combination, these results have negative implications for postural stability in older adults when the sensory environment is in motion. They downweight vision gains less fully (in the absolute sense) than young adults. In addition, fall-prone older adults downweight more slowly than healthy older and young adults. Remaining relatively more coupled to a non-stationary source of sensory information may compromise MSR estimation processes and postural stability over a longer period of time.

4.3 Differing older adult responses to high versus low amplitude visual motion help explain discrepancies between previous studies:

In a previous study of MSR with healthy young, healthy older and fall-prone older adults (Allison et al, in press) we found no between-group differences in the absolute levels of vision and touch gains, or in the pattern of gain adaptation between conditions with varied visual and touch motion amplitude stimuli. This finding was surprising given several previous studies whose results demonstrated clear differences between young and older adults that were attributed to impaired or slowed MSR (Stelmach et al, 1989; Teasdale et al, 1991a, 1991b, 1993; Hay et al, 1996). In our previous study, subjects were exposed to visual motion stimuli amplitudes of 2, 4, and 8 mm, and gains were averaged across cycles in two minute trials. We subsequently hypothesized that we might have obtained discrepant results because we used long trials while prior studies typically used much shorter trials: if healthy and/or fall-prone older adults reweight more slowly than

young adults, between group differences might be more apparent in studies with short trials and obscured in our study with long trials. In hindsight, we also note now that many prior studies using visual motion to explore age-related changes in MSR typically used larger amplitude visual stimuli (c.f., Sundermeier et al, 1996; Simoneau et al, 1999).

Results from the current study, in which responses to small and large amplitude visual motion stimuli were compared and gains were examined over shorter timeframes, explain why our previous results were inconsistent with prior studies. First, healthy and fall-prone older adults responded differently to high versus low amplitude visual motion. Results from the current study in conditions where the visual motion amplitude is high (higher absolute gains in older vs. younger adults) are consistent with prior investigations, while results in conditions where the visual motion amplitude is low (similar absolute gains in older vs. younger adults) are not. Second, when visual motion became less stationary, young adults downweighted vision rapidly and fully in five seconds or less, while healthy and especially fall-prone older adults continued to gradually downweight over a prolonged period of time and never downweighted to the extent that young adults did. So studies using shorter trials (typically on the order of 10 to 30 seconds) may have captured completed re-weighting in young adults but incomplete re-weighting in healthy and fall-prone older adults.

Teasdale et al (1991b) and Hay et al (1996) offered convincing evidence that central sensory processing was slowed or impaired in older versus younger adults. They demonstrated increased instability in older but not younger adults when reliable sensory information was restored after a period of withdrawal. When a relatively more reliable source of sensory information becomes available, estimation processes and postural

stability are optimized by upweighting (increasing gains to) that information source. In their studies, then, upweighting appeared to be intact in young subjects but deficient in older subjects. In contrast, we did not find between group-differences in absolute levels of gain, or in either rapid or prolonged adaptation, when upweighting was required after a switch from high to low amplitude visual motion stimuli. However, our subjects were not exposed to conditions where vision was withdrawn and then reinserted. It may be that the variations in visual motion amplitude we provided were less demanding on the upweighting process than having to upweight from (essentially) zero when vision is absent to maximal gain when stationary, reliable visual information is reinserted.

4.4 Position variability and velocity variability

Position and velocity variability measures provide information about residual postural sway responses at frequencies other than the visual motion stimulus frequency. When visual information is relatively more stationary (low amplitude), gains increase and sway variability decreases. When visual information is relatively less stationary (high amplitude), the converse is true. We thus expected higher variability in the Pre-High and Post-High conditions, and lower variability in the Pre-Low and Post-Low conditions. Results from previous studies (Jeka et al, in press; Allison et al, in press) led us to expect that differences would be more apparent in velocity versus position variability. Our results are consistent with this expectation. We found no significant differences in position variability between groups or conditions, and no group-by-condition interaction.

Velocity variability was significantly lower in the Post-Low condition than in all other conditions when gains were increasing but was not lower in the Pre-Low condition

when gains were decreasing. There were no between-group differences in velocity variability in any condition, meaning that the pattern of residual postural sway response was similar in all three groups. This finding is surprising. When stimulus amplitude was high, absolute levels of gain were higher and prolonged downweighting was slower in healthy and fall-prone older adults compared to the young: we might have expected higher variability in the older adults. However, in this circumstance gains were decreasing. In this study we found that velocity variability was lower only in the condition where gains were increasing (Post-Low).

Summary and Conclusions

Multi-sensory re-weighting is impaired but not absent in healthy and fall-prone older adults; this raises the possibility that MSR processes might be preserved or remediated through sensory-challenge activities or exercise. Higher vision gains in both older groups as compared to the young may reflect a compromised ability to upweight somatosensory and/or vestibular inputs, though this explanation remains to be tested directly. The combination of higher vision gains and slower downweighting in fall-prone as compared to healthy older adults may represent deficits that contribute to postural instability and falls.

Future research should continue to explore the mechanisms of postural instability in fall-prone older adults, including but not limited to MSR. In particular, visual dependence or visual sensitivity in older adults warrants further investigation. Ideally, research designs that permit all three sensory sources to be simultaneously manipulated

and measured will be developed. This would permit a greater understanding of the adaptive relationships between sensory sources in response to changes in the environment, as well as a direct test of the hypothesis that higher visual gains are the result of an inability to upweight somatosensory and/or vestibular inputs. In the future it will also be important to study the degree to which MSR deficits impact postural stability and fall-risk relative to other co-existing impairments in postural control systems such as peripheral sensory loss, muscle weakness, etc. Sensory-challenge exercise programs that can be easily performed in clinical and/or community settings need to be developed and formally tested for their effects. It appears that such exercises result in lowered vision gains (Allison et al, submitted), but it would also be important to test whether an increase in the rate of re-weighting occurs. Assuming exercise has these effects, we would also like to know (1) how much improvement in both the level of gain and rate of gain change might be expected from such an exercise intervention, and (2) whether this amount of improvement at the impairment level translates into any real benefit re: improved balance, reduced risk of falls, and increased function.

Figure Captions

FIGURE 4-1

Exemplar postural sway trajectory.

The medial-lateral center of mass trajectory (solid line) from one fall-prone older adult in the previous study is shown with the visual stimulus (dotted line) oscillating at 0.2 Hz with an amplitude of 8 mm. Note that the amplitude of the postural sway is initially high and then gradually dampens over the 60-s. time segment.

FIGURE 4-2

Experimental set-up.

A subject is shown in a standardized stance on the forceplate facing the front screen of the computer generated visual display.

FIGURE 4-3

Visual motion stimuli

The visual motion stimulus oscillated at 0.4Hz in both conditions. In the High-to-Low condition (top), stimulus amplitude was 12 mm for 60-s, then switched to 3 mm for 120-s. In the Low-to-High condition (bottom), stimulus amplitude was 3 mm for 60-s, then switched to 12 mm for 120-s.

FIGURE 4-4 A – C

Time series.

Three exemplar time series, each from one trial of the 60-s pre-switch time segment in the Low-to-High condition, are shown from (A) a fall-prone older adult, (B) a healthy older adult, and (C) a healthy young adult. Each graph shows the time series of visual display motion at 0.4 Hz and the anterior-posterior center of mass displacement. In these exemplars, visual motion amplitude is 3mm (low amplitude).

FIGURE 4-5 A – C

Amplitude spectra.

Three exemplar amplitude spectra, each for five-trial averaged data from the 120-s post-switch time segment in the High-to-Low condition, for the same three subjects shown in Fig. 4-4: (A) fall-prone older adult, (B) healthy older adult, and (C) healthy young adult. For each graph, top lines represent the averaged anterior-posterior center of mass and bottom lines represent the visual stimulus. In these exemplars, visual motion amplitude is 3mm (low amplitude). Note that almost all the power of the visual stimulus is at the driving frequency of 0.4 Hz. Each subject shows a clear response to the stimulus at this frequency.

FIGURE 4-6 A – B

Group mean gain results.

The mean vision gains for each group, with regression lines indicating the prolonged rates of gain change in each condition. Group differences in the absolute levels of gain are most apparent when the visual motion stimulus amplitude is high.

FIGURE 4-7 A – B

Group mean phase results.

The mean vision phases for each group, with regression lines indicating the prolonged rates of gain change in each condition.

FIGURE 4-8 A – B

Group mean position variability results.

The mean (\pm S.E.) position variability for each group at each 60 second time segment in each condition.

FIGURE 4-9 A – B

Group mean velocity variability results.

The mean (\pm S.E.) velocity variability for each group at each 60 second time segment in each condition. Note that when the visual motion stimulus amplitude is high [and gains are low], velocity variability is high, and vice versa.

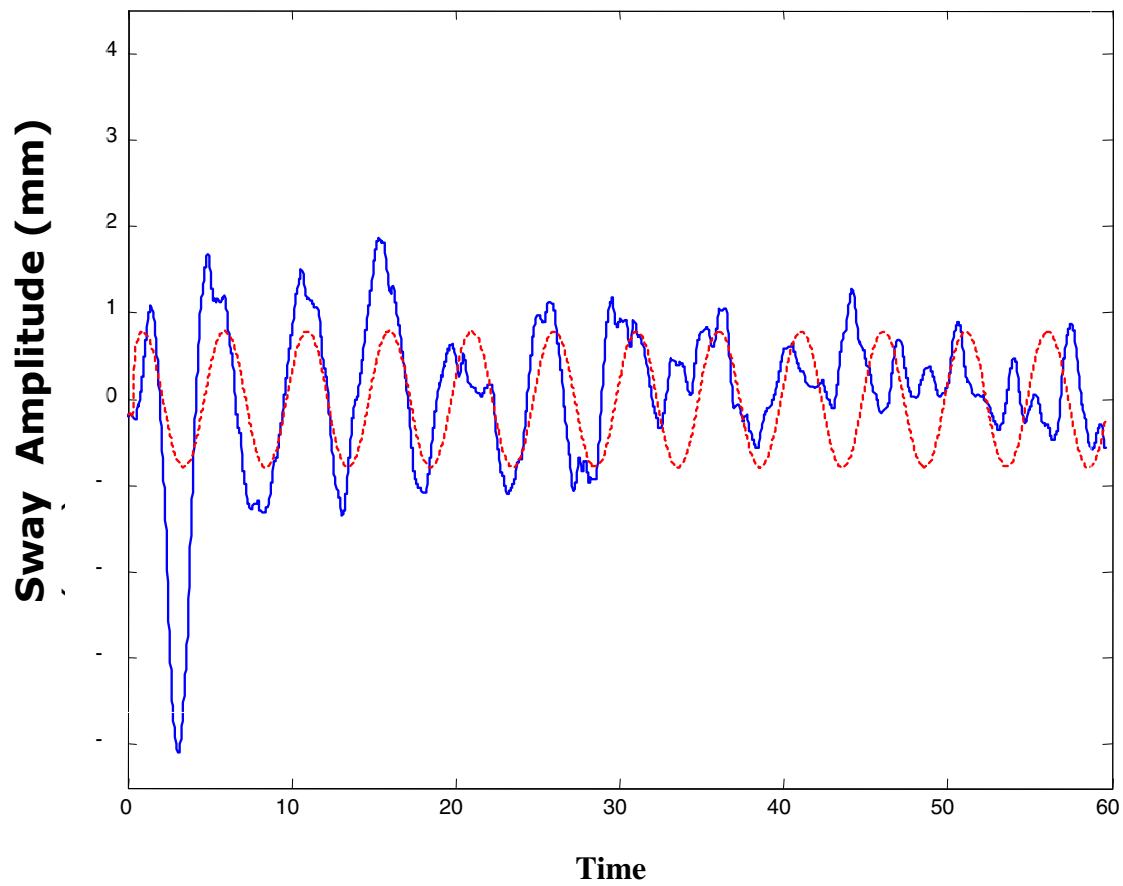


Figure 4-1: Exemplar COM sway trajectory



Figure 4-2: Experimental set up

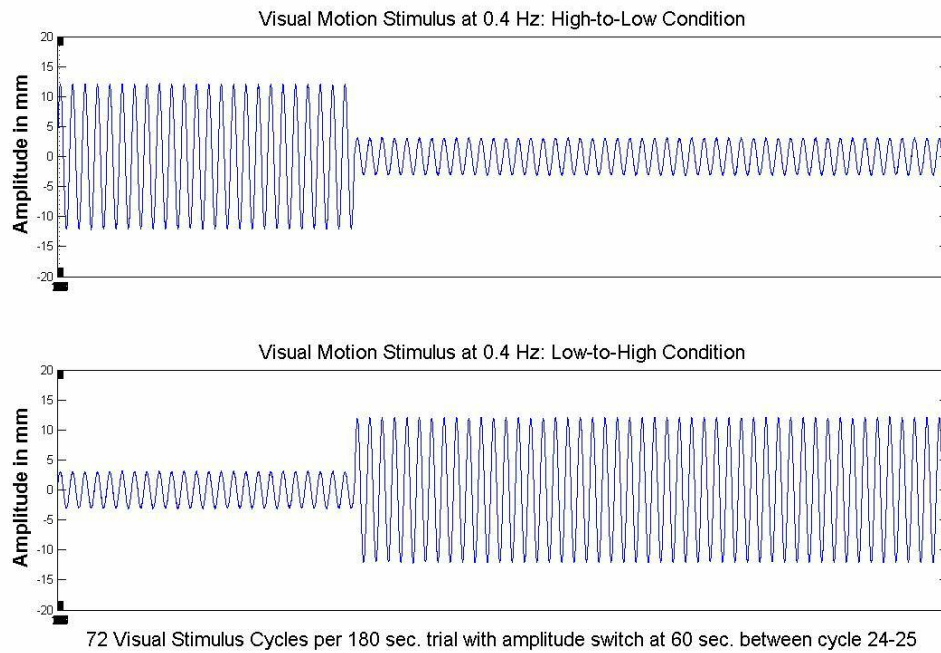


Figure 4-3: Visual motion amplitude stimuli

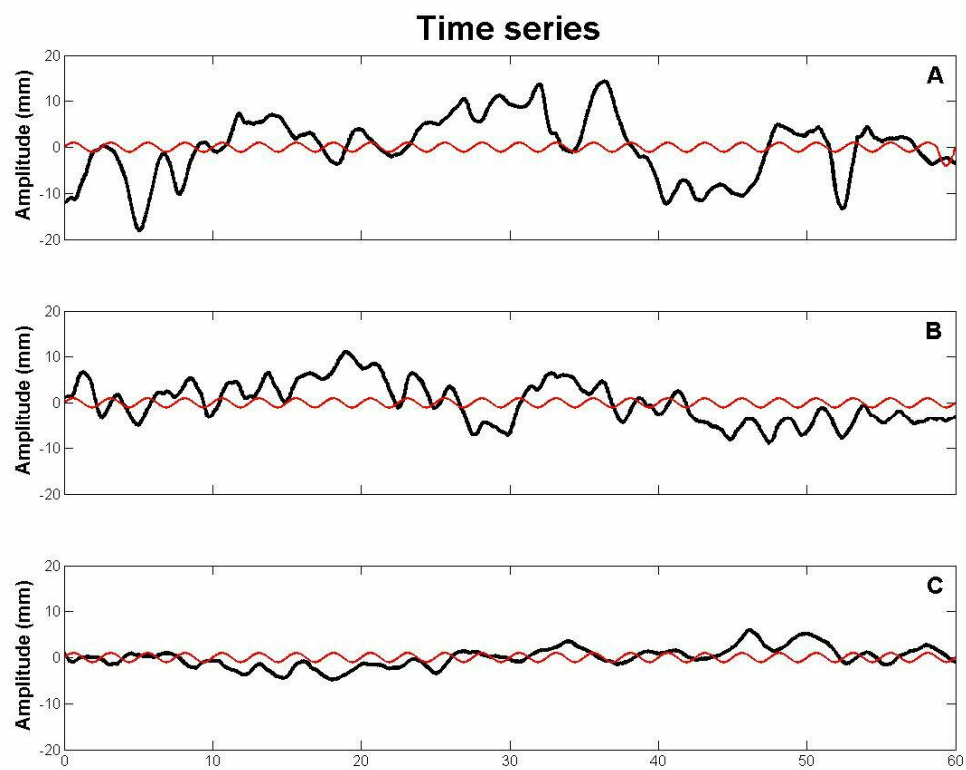


Figure 4-4 A-C: Exemplar times series

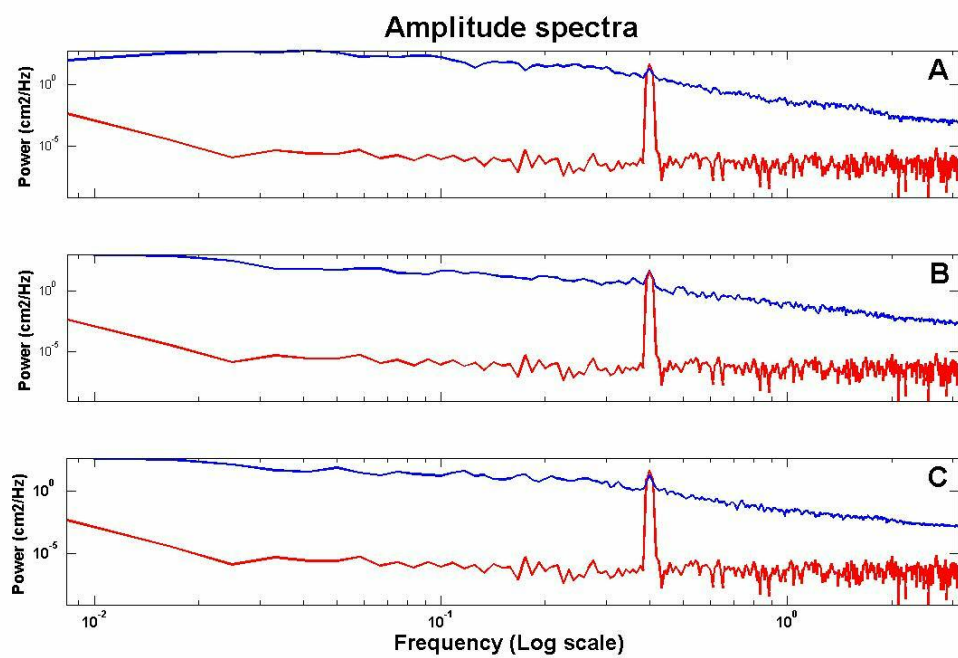


Figure 4-5 A-C: Exemplar amplitude spectra

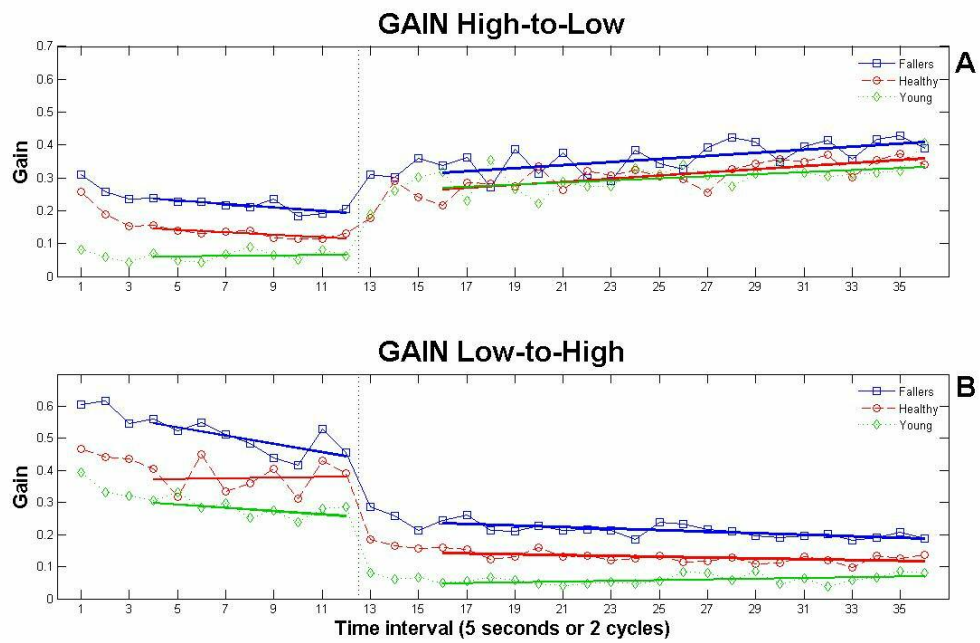


Figure 4-6 A-B: Vision gains with regression lines

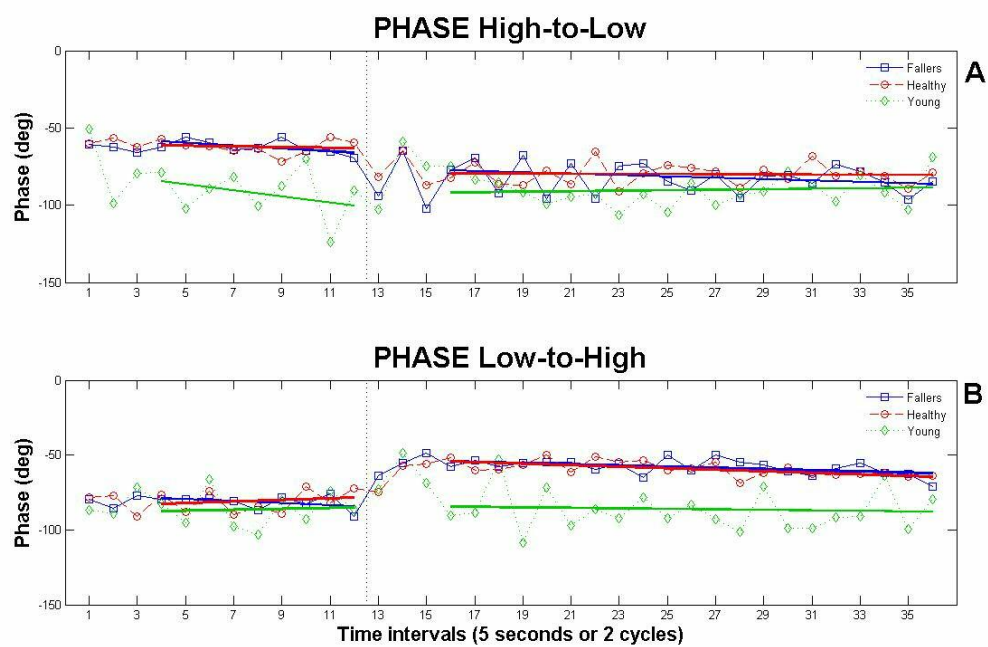


Figure 4-7 A-B: Vision phases with regression lines

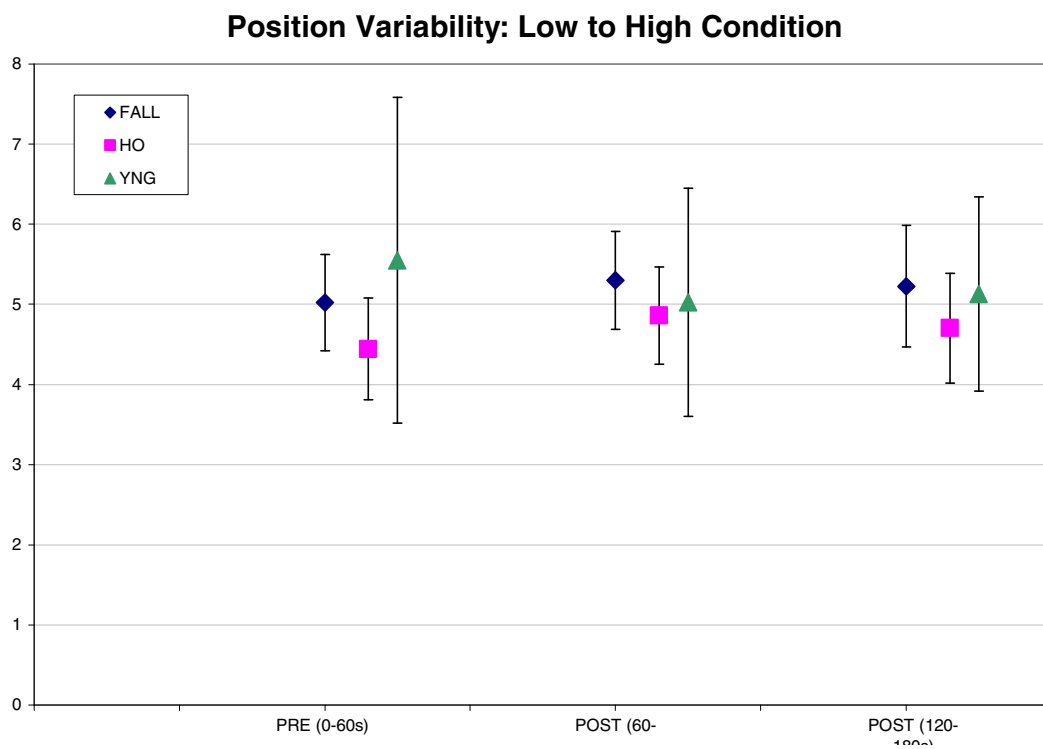
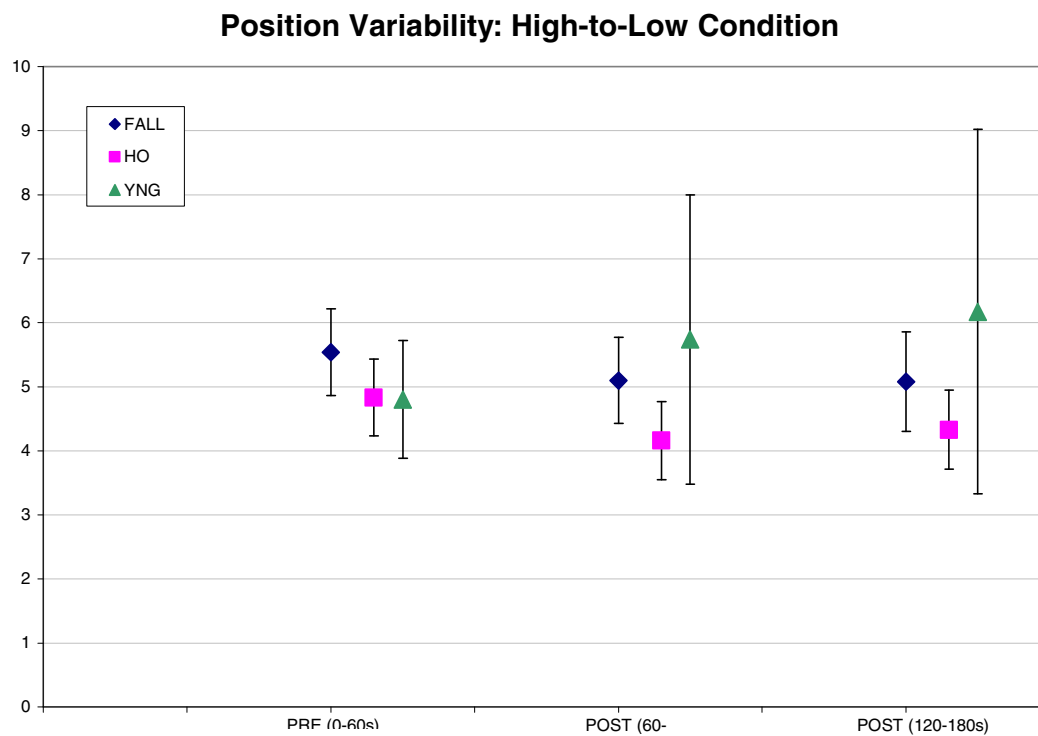


Figure 4-8 A-B: Position Variability

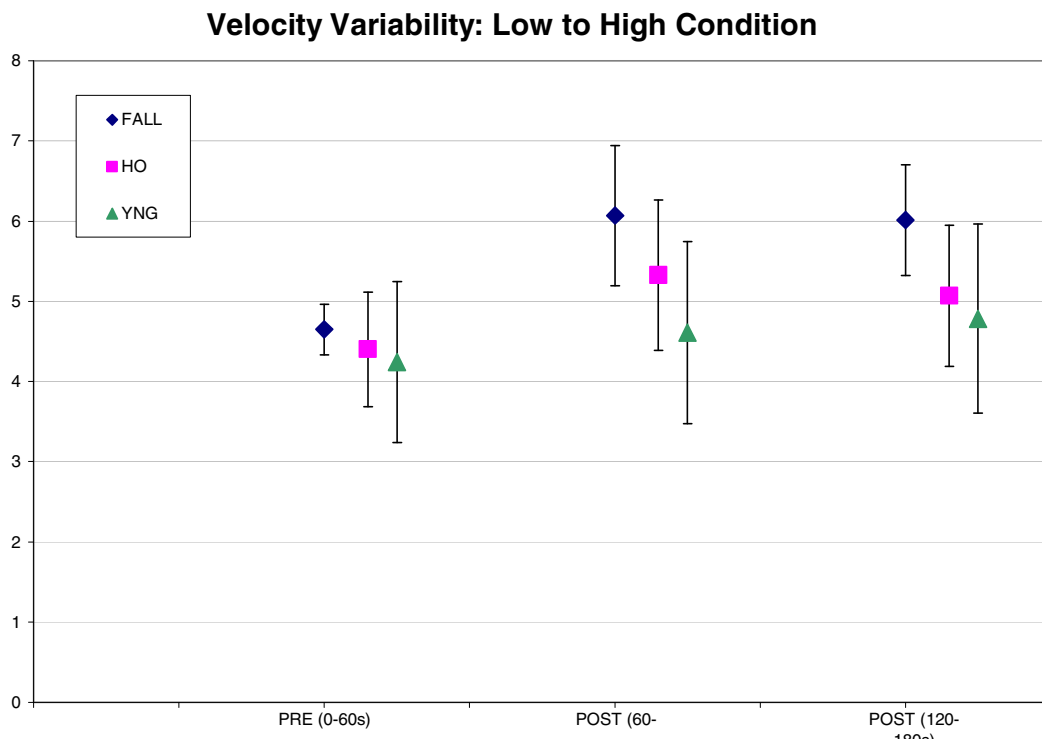
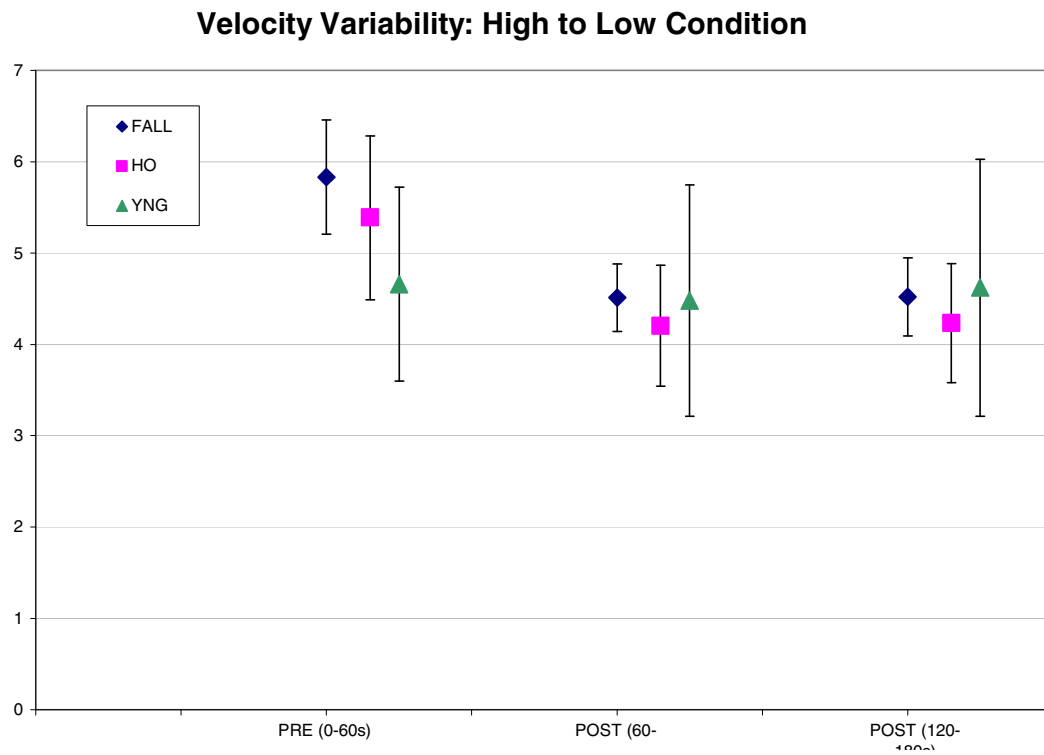


Figure 4-9 A-B: Velocity Variability

Chapter 5: Summary and Future Directions

Falls are a serious problem for a large subset of older adults. Because the population of adults over 65 is growing rapidly, the number of annual fall-related injuries and deaths is also bound to grow. As the extent and severity of the problem has been revealed, determined efforts on the part of the scientific and health care communities are being made to address the issue. Researchers who study postural control are exploring age-related changes in the sensori-motor systems that contribute to dynamic human equilibrium, and investigating the subtle pathological differences between older adults who fall and those who do not. Health care providers from numerous disciplines are working to apply this expanding knowledge to clinical practice. Methods to identify, evaluate and treat fall-prone older adults, and prevention strategies for healthy older adults to preclude their need for such services, are being developed and implemented.

An accumulation of results from numerous studies makes clear that imbalance (postural instability or dyscontrol) is the primary intrinsic risk factor for falls. Yet because postural control is a multi-dimensional process, the study of age and pathology-related changes in postural control is complex. Any one or more of the peripheral and central sensory and motor system components may be involved, with the normal flexibility and adaptability of the neuromusculoskeletal system becoming progressively more compromised as the number of impaired components increases. Further complicating the picture is the fact that the older adult population is highly variable in terms of health status and physical functioning. Even among the subset of fall-prone older adults there is great heterogeneity; different older adult fallers present with different sets of postural control system impairments.

One component of the postural control system that is believed to change with age and to be critically deficient in the fall-prone elderly is multi-sensory re-weighting (MSR). Multi-sensory re-weighting is an adaptive central sensory process that prioritizes the visual, somatosensory and vestibular inputs that provide the most available, accurate and reliable information to the nervous system. This on-going process permits the estimation of body position and motion in space to be continually updated as conditions in the sensory environment change. Deficiencies in MSR would impair such estimation, and consequently might alter perceptual recognition of the need for balance responses and the accuracy of neural commands to the musculature for the generation of effective balance maintenance and recovery strategies. In this way MSR deficits may contribute to imbalance and falls in older adults.

The study of age-related changes in MSR and the potential contribution of MSR deficits to falls in older adults has been the focus of this dissertation. A review of previous research on sensory organization/integration in the elderly to identify knowledge gaps and the development of more specific quantitative postural control measures motivated our approach. In these studies we have employed three groups of subjects, healthy young adults, healthy older adults, and fall-prone older adults. Comparison of healthy young and healthy older adults allows the study of changes in MSR that are due primarily to aging. Comparison of healthy and fall-prone older adults permits the examination of changes in MSR that are not solely due to aging and may represent pathology and heightened fall risk. To significantly reduce the possibility that peripheral sensory loss might affect central sensory processing and thus confound our results, we screened each older adult to ensure that they had sufficiently intact peripheral sensation

prior to inclusion in the study. To diminish the possibility that balancing on an unstable surface might impose biomechanical challenges that would affect postural sway in the more frail older subjects, somatosensory inputs were manipulated using fingertip touch. In each study we used recently developed quantitative measures that have permitted a closer inspection of MSR mechanisms than was possible in previous studies of sensory organization/integration in the elderly. For example, provision of visual and touch stimuli at different frequencies and analysis of postural responses in the frequency versus the time domain made it possible to separately quantify responses, and explore intra- and inter-sensory re-weighting, to the visual and touch stimuli. Likewise, additional analysis of position and velocity variability made it possible to discriminate changes in the postural response attributable to the altered sensory stimuli versus changes in overall [residual] postural sway that could be influenced by several other non-sensory factors. Taken together, the results of the three reported studies both offer support for and stand in contrast to the findings of prior research.

In the first study we investigated whether or not MSR of vision and touch was impaired in healthy and fall-prone older adults compared to healthy young adults. Standing subjects were exposed to simultaneous medio-lateral oscillatory visual and fingertip touch inputs at varying relative amplitudes. Both healthy and fall-prone older adults demonstrated the same pattern of adaptive gain changes as healthy young adults. Like the young adults, both older groups displayed clear evidence of intra- and inter-sensory re-weighting to both vision and touch motion stimuli. No group differences in the overall levels of vision and touch gain were found. There were no between group differences in position or velocity variability. For all groups, velocity variability was

higher in conditions requiring more substantial inter-sensory re-weighting. These results suggest that, for small amplitude vision and touch stimuli, the central sensory re-weighting adaptation process remains intact in healthy and fall-prone older adults with sufficiently intact peripheral sensation. Further, at all ages, re-weighting is an imperfect compensatory process that cannot completely make up for lost information.

In the second study we examined the effect of sensory challenge balance exercises on laboratory measures of MSR (e.g., gain, phase, etc.) and clinical measures of balance in the fall-prone older adults that had served in the first study. Twenty eight older adults completed 16 individual 45-minute exercise sessions. Two baseline and one post-intervention test sessions were conducted. Clinical measures included the Activities-specific Balance Confidence Scale [ABC], Berg Balance Scale [BBS], Sensory Organization Test[®] [SOT], Limits of Stability[®] test [LOS], and lower extremity strength and range of motion. Laboratory measures included center of mass [COM] gain and phase to dynamic vision and touch stimuli, and COM position and velocity variability. Following the intervention a significant decline in overall touch gain and a nearly significant decline in overall vision gain were observed. No changes in position or velocity variability were seen. Expected improvements in SOT scores, and unexpected improvements in scores on the BBS, LOS and lower extremity strength tests, were observed. These findings confirm intact MSR pre- and post-intervention, and offer support for improved perceptual discrimination and dissociation of self-motion versus environmental motion after exercise designed to specifically target multi-sensory integration mechanisms. The positive effects of the targeted intervention generalized to

other clinical balance measures. Lowered susceptibility to potentially perturbing environmental motion stimuli may improve postural control and lower the risk of falls.

The primary outcome of the first study, that healthy and even fall-prone older adults displayed intact MSR, was in contrast to the results from numerous prior studies that indicated deficient MSR in these populations. We hypothesized that, if MSR in healthy and fall-prone older adults was intact but occurred more slowly than in young adults, then the long trials in our study may have permitted these older subjects to re-weight more fully compared to the (relatively) short trials common to previous investigations. (In hindsight we also can see that the small amplitude stimuli used in our first two studies were unlike the larger and faster perturbations typically used in other studies.) This possibility, that MSR processing was slowed rather than deficient, motivated our third study. There we explored the dynamics of re-weighting to changes in visual motion stimuli in healthy young, healthy older, and fall-prone older adults. Subjects were exposed to two visual motion conditions in which the amplitude of visual motion switched from low-to-high or high-to-low. Postural sway responses to these changes in the visual environment were analyzed using gain, phase, position variability and velocity variability. Absolute levels of gain at, and the rates of adaptive gain change across, selected time intervals were compared between groups. Both rapid and prolonged rates of changes were examined. Absolute levels of gain were consistently higher in both older groups compared to the young when the visual motion stimulus amplitude was high. Gains were frequently higher in the fall-prone versus healthy older and young adults when the visual motion stimulus amplitude was low. All three groups demonstrated adaptive sensory re-weighting reflected by gain changes following stimulus

amplitude changes. All three groups showed equally rapid re-weighting at the time of the amplitude switch (within the five-second resolution window). Between group differences were apparent in the rates of prolonged change in gain. Compared to young adults who usually did not re-weight further after the initial rapid adaptation, both older groups demonstrated continued gradual changes in gain over time periods of 105 seconds. When the stimulus amplitude was high, both older adult groups demonstrated slower prolonged adaptation rates than the young. Rates of prolonged adaptation were not different between the older groups and the young when the stimulus amplitude was low. No differences in position variability were observed before and after the switch for either condition. Velocity variability was higher when stimulus amplitude was high versus low, in both conditions, again illustrating that compensatory re-weighting cannot fully offset the effects of reduced sensory information. Rapid re-weighting when the stimulus amplitude suddenly increases is necessary to prevent instability. All three groups demonstrated functionally adaptive responses. However, down-weighting to high amplitude stimuli is slower and to a lesser extent in healthy and fall-prone older adults compared to healthy young adults. With low-amplitude stimuli, fall-prone older adults continued to have the highest absolute levels of gain and slowest rates of prolonged change compared to healthy older and young adults.

While the studies reported herein have added valuable knowledge to the understanding of age and pathology related changes in MSR, the area remains understudied. Much more work is required before substantial application of this information can be made confidently to further address the serious problems related to imbalance and falls in older adults. Future research in this area should place increased

emphasis on the each of the multiple hypothesized mechanisms of postural dyscontrol, their relationship(s) to each other, and the combined effect of multiple deficient mechanisms which is likely the case in most fall-prone older adults. For example, within the MSR process, are heightened gains to visual motion stimuli accompanied by diminished gains to vestibular and somatosensory stimuli? If so, is the increase in vision gains a secondary compensatory strategy to counter an inability to upweight vestibular and somatosensory inputs? If not, what purpose does such a neural strategy serve? Further, what effect do deficits in estimation processes such as MSR have on control mechanisms such as stiffness and damping? Do changes in MSR affect automatic postural responses, anticipatory postural adjustments or voluntary COG control?

Future studies also should include specific, quantifiable measures of the postural control mechanisms under investigation in addition to behavioral outcome measures of balance, and should relate the findings from each level of study. For example, what is the relationship between higher gains to dynamic sensory stimuli and scores on behavioral indices of balance such as the Berg Balance Scale and the Timed Get Up and Go test? If, following intervention, gains are lowered and/or the rate of change in gain is increased, are improvements noted in behavioral measures of balance that reflect a reduced risk of falls? Until several such studies are performed, our understanding of the relationship between MSR and postural response behavior remains obscured. Consequently, our ability to devise interventions that specifically target deficient mechanisms is also hampered.

From the results of the three studies reported above, additional related issues and questions arise. The consistent finding that MSR is not absent in healthy or fall-prone

older adults is encouraging; MSR appears to be a process that is relatively preserved in aging and as such might be intentionally facilitated to help compensate for other postural control processes that are more affected by age. However, there are alterations in the MSR process, particularly the existence of higher gains to high amplitude visual motion stimuli, and a diminished ability to down-weight these high gains as rapidly and completely as young adults do. Sensory challenge balance exercises appear to reduce overall gains to dynamic vision and touch stimuli, but there is much more to learn. We measured only vision and touch gains, but not vestibular gains, so we do not know if the post-training decrease in vision and touch gains was perhaps in response to an improved ability to up-weight vestibular inputs or rather an improved ability to scale the extent of coupling to sensory stimuli. To probe this question, future studies should ideally manipulate and measure responses to all three sensory modalities simultaneously.

Following the sensory challenge balance exercise program, we measured overall levels of gain but did not obtain pre- and post-measures of the rate of adaptive gain change, so we do not know if these exercises had any positive effect on the dynamics of re-weighting. Further, we did not use any high-amplitude stimuli in our pre- and post-testing.

Responses to high-amplitude stimuli are more deficient in the older adults and may be more resistant to change. Thus future studies of the effect of sensory challenge balance exercises on MSR should include a wider range of stimulus amplitudes and should also measure the rate of adaptive gain change before and after intervention. In our study of the dynamics of sensory re-weighting, we manipulated and measured the response to only visual inputs. Additional studies of the dynamics of re-weighting should manipulate and measure the responses to somatosensory and vestibular inputs as well, and subsequently

manipulate and measure the responses to combinations of stimuli (e.g., vision and touch, etc.) to explore the dynamics of MSR.

An area of great interest that deserves further exploration is that of ‘visual dependence’ or ‘visual sensitivity’ in healthy and fall-prone older adults. The observation that healthy and especially fall-prone older adults respond more strongly to environmental visual motion has been made previously by other researchers and this behavior was evident in our most recent study using higher amplitude visual motion stimuli. A frequent explanation is that peripheral somatosensory loss precludes sensory coupling to somatosensory stimuli, hence a compensatory increase in the use of visual inputs. However, we observed this response in healthy and fall-prone older adults without clinically detectable peripheral somatosensory loss, as have others, and this finding does not support the aforementioned explanation. We further found that following a sensory challenge exercise program, both vision and touch gains were reduced. If high vision gains were linked to poor use of touch cues, then the training program should have produced a decline in vision gains and an increase in touch gains, and this was not the case. It is possible that heightened vision gains are instead a compensation for peripheral vestibular loss or poor use of vestibular inputs; if this were the case and the use of vestibular inputs was improved post-training, that might better explain the joint decrease in both vision and touch gains. As yet careful and thorough measures of vestibular function, and the measurement of vestibular gains, have not been employed in studies of MSR or sensory challenge exercise interventions with older adults. To address the issue of visual dependence in healthy and fall prone older adults, future studies should explore postural responses to a broad range of visual motion stimulus amplitudes, use high-level

diagnostic tests to evaluate the peripheral visual, somatosensory and vestibular systems, and perhaps include psychometric testing of motion perception for the three modalities.

Lastly, more studies of the effects of sensory-challenge balance exercises are warranted. There appears to be a relationship between the performance of sensory-challenge balance exercises and improvement on multiple measures of postural control, but more work needs to be done to establish a causal relationship, if one exists. Clinically feasible sensory-challenge balance exercise programs need to be developed and tested for efficacy. It is not expected that clinicians would use a sensory-challenge balance exercise program as the sole intervention, yet the relative value of such exercises compared to other forms of exercise such as strengthening or Tai Chi should be explored. It is more probable that sensory-challenge balance exercises would be included as one component of a multi-dimensional exercise program, and in this context the relative ‘added value’ of such exercises in combination with other forms of exercise should be established. Future studies should also ideally determine how much improvement [in the overall level of gain and in the rate of gain change] may be expected following participation in a sensory-challenge balance exercise program, and if this amount of improvement is sufficient (alone or in combination with other exercises) to improve balance, reduce the risk and incidence of falls, and increase functional mobility in fall-prone older adults.

Much of the information gained from the studies described here would not have been obtained using more traditional methods of investigation. The tools and techniques to investigate postural control and MSR continue to advance. As they do, so will our level of understanding and our ability to apply this knowledge toward preventive care and the enhancement of successful aging process for older adults.

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